AN ABSTRACT OF THE DISSERTATION OF

Seung uk Ko for the degree of Doctor of Philosophy in Mechanical Engineering presented on October 16, 2006.

Title: Human Gait Analysis by Gait Pattern Measurement and Forward Dynamic Model Combined with Non Linear Feedback Control.

Abstract approved:

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Normal gait of human is believed to be an adapted motor control result of optimal behavior of individual. Normal gait would be altered by aging or injury without being perceived. Subtle changes in gait which is not easy to detect, can cause many serious problems including joint pains and even unexpected falls. Finding out the gait parameters which can be an identifier of abnormal gait is main goal of this study. Gait parameters were compared between elderly participants who had different falling experiences. Relatively easy temporal and spatial gait parameters were measured and used for side way direction faller identification. As a new method of calculating kinetics, forward dynamics method combined with linearized feedback control (FDMLFC) was developed and used for the consumed mechanical work calculation.

Human Gait Analysis by Gait Pattern Measurement and Forward Dynamic Model

Combined with Non Linear Feedback Control

by

Seung uk Ko

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I understand that my dissertation will become part of the permanent collection of Oregon State University libraries. My Signature below authorizes release of my dissertation to any reader upon request.

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GENERAL INTRODUCTION

The human gait is inherently unstable, essentially consisting of a series of interrupted falls. Normal gait patterns of healthy adults can be considered as the result of optimized motor control. The optimized normal gait pattern can be impaired by the aging process. Impaired gait patterns can be affected by deformity, muscle weakness, pain, and abnormal motor control (including sensory loss and spasticity). The main goal of this study is to find out the modified (without being perceived) gait patterns which could have been the reasons for past falls and can cause the future falls of older adults. Two studies (the first and third ones) provided here were based on two different human subject gait tests. Participants who volunteered for these two studies were over 70 years old and can walk without any walking assistance devices. None reported any history of neurological or orthopedic problems, but each person has different types of falling experiences, including no falling experience. Various gait patterns in the parameters of kinematics and kinetics were compared based on the falling experiences which participants had reported.

Hip fractures are one of the most debilitating and costly fall-related injuries suffered by older adults. Of all falls, those to the side with impact on the hip increase the risk more compared to falls in any other direction. The first part of this work focused on finding the identifier which can distinguish the sideways fallers from the other direction (forward or backward) fallers based on spatial and temporal gait parameters. One hundred thirty nine healthy volunteer individuals participated in this first study. Over the two-year surveillance period, 29 subjects fell to the side and 64 subjects fell in “other” directions (forward, backward, straight down). Gait parameter comparisons were made between the groups to characterize the gait pattern of sideways fallers. Significant difference was found in the gait parameter of ‘stride width’. Sideways fallers showed narrower stride width compared with other direction fallers.

The new method for three dimensional kinetic analyses was developed in the second part of this study. Forward dynamics method combined with linearized feedback control system (FDMLFC) was introduced. FDMLFC can be used as an alternative of the traditional inverse dynamics method without error propagation for more applications with more flexibility. The applied feedback control in the FDMLFC method solves the “ill posed problem” which traditional forward dynamics method should deal with.

In the third part of this study, comparisons in gait parameters such as spatial, temporal gait parameters, and the consumed mechanical work from the lower extremities in sagittal and frontal planes were made between the groups to find the gait pattern characteristics of the healthy elderly fallers, which is significantly different from the healthy elderly non-fallers. Twenty-eight healthy elderly volunteer individuals participated in this study. Twelve of them were classified as fallers based on their reports about falling experiences within the last twelve months. Normal gait tests were performed by using a three dimensional motion capture system, and the gait parameters were calibrated and calculated. The kinetics required during gait tests were obtained from the FDMLFC method, and the consumed mechanical works were calculated. Gait parameter comparisons between the groups revealed the gait pattern characteristics of healthy elderly fallers. Fallers showed narrower stride width with larger standard deviation in mechanical work generation of knee extensor compared with non-fallers.

STRIDE WIDTH DISCRIMINATES GAIT OF SIDE-FALLERS COMPARED TO OTHER-DIRECTION FALLERS DURING OVERGROUND WALKING

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ABSTRACT

Our purpose was to identify differences in gait characteristics between older fallers with a tendency to fall sideways compared to those who do not fall to the side. **Methods:** We conducted a prospective, case-control study of ambulatory adults over 70 years residing in retirement communities. Measurements included spatial and temporal gait parameters and prospective fall surveillance. **Results:** Twenty-nine subjects fell to the side and 64 subjects fell in "other" directions (forward, backward, straight down). Forty-six subjects experienced no falls. Side-fallers exhibited narrower stride widths compared to other-directed fallers and a stepwise discriminant analysis correctly classified 67% of side-fallers and other-directed fallers using only stride width. **Discussion:** This study suggests that side-fallers, who have narrower stride widths compared to those who fall in other directions, may not be adapting their gait to compensate for lateral instability. More research is needed to determine whether narrow gait contributes to unstable walking patterns.

INTRODUCTION

Falls by older adults are the leading cause of injuries and hospital admissions for trauma in the U.S. Hip fractures are one of the most debilitating and costly fall-related injuries suffered by older adults. Of approximately 350,000 hip fractures annually, over 90% result from a fall.[1, 2] Of all falls, those to the side with impact on the hip increase the risk for fracture 20 to 50 times compared to falls in any other direction.[3, 4, 5] In spite of the widely appreciated magnitude of this problem, very little is understood about the mechanisms contributing to falls in the sideways direction.

Most injurious falls in older adults occur while walking.[6, 7] Human walking is inherently unstable, essentially consisting of a series of interrupted falls. Stability while walking is particularly challenging because it requires controlling the motion of the whole body center of mass within a constantly changing and moving base of support provided by the feet. To date, most studies of walking and fall risk have focused on temporal aspects of gait (e.g., swing time, stance time and gait speed), and stability in the antero-posterior direction (e.g., step length, stride length).[8, 9, 10, 11, 12] Older adults alter gait patterns by reducing walking speed, stride length, and stride frequency, and they exhibit greater stride to stride variability compared to younger individuals.[11, 13] These patterns are exaggerated among individuals with a tendency to fall.[9, 11, 13, 14, 15] Almost no attention has been given to the problem of side-falls, or to deficits in lower extremity function and the control of medio-lateral stability[16, 17, 18, 19, 20] that could predispose participants to high risk falls to the side. Work from our laboratory has shown that performance factors related to medio-lateral stability predict individuals that fall to the side versus "other-directed" fallers[21] and data from passive walking models suggest that medio-lateral foot placement is an effective strategy to stabilize lateral balance.[22] In a study comparing older fallers to younger individuals and older non-fallers, the older fallers demonstrated significantly greater sideways body motion toward the stepping side at first-step foot contact and a more laterally directed foot placement.[19] A study by Donelan found that lateral instability affects the choice of preferred stride width and that there was a 43% reduction in preferred stride width with external stabilization of the pelvis during gait.[23] This implies that the less stable an individual is in the medio-lateral plane, the more likely they might be to adopt a wider gait pattern. However, visual observations of side-fallers in our laboratory suggested they walk with a narrower gait compared to their peers who also experienced falls, but none to the side. To date, nobody has examined medio-lateral stability during gait in a population of individuals with a history of losing balance and falling in the sideways direction.

The purpose of this study was to examine spatial and temporal aspects of gait in a population of older side-fallers compared to individuals who fell in "other" directions (forward, backward, straight down) to determine if gait parameters discriminate fall direction. We hypothesized that side-fallers would walk with narrower strides and exhibit greater stride-width variability and that these variables would distinguish side-fallers from those who fell in other directions.

METHOD

DESCREPTION OF THE DATA SET

Data presented here come from individuals who had been undergoing fall surveillance for one year as part of another study investigating side-fall incidence in older adults.[24] All participants were recruited from retirement facilities in and around Oregon's Mid-Willamette Valley.

The subject group for the present study consisted of 139 healthy volunteer individuals (109 women, 32 men) ranging in age from 72 to 94 years (mean 83, SD 5.6). None reported any history of neurological or orthopedic problems likely to affect mobility or balance and all subjects were able to ambulate without the use of a walking aid (e.g., cane, walker). Additionally, none had any physical or cognitive conditions that precluded them from understanding or safely participating in the physical tests. Subjects were excluded from the study if they were unable to fill out health history and screening questionnaires, were taking any psychoactive medications[25, 26], or if they were unable to report their falls incidence reliably over the two-year observation period. All subjects completed health history questionnaires and underwent gait and mobility testing at baseline. During the previous year, as part of a separate study, participants were instructed how to track any and all falls that they might experience over the next year using a falls diary. Participants were given new falls diaries and asked to continue logging all falls for the subsequent year. Fall status for the present study was determined using surveillance data from the year prior to gait testing and the year following gait testing. The Institutional Human Subjects Review Board gave approval for this study and informed consent was obtained from all subjects prior to their participation.

MEASURES

Gait Analysis. Data were collected using an electronic walkway system (GaitRite Electronic Walkway System, CIR Systems, Clifton, NJ) to determine gait characteristics of participants. The walkway is a portable mat that can be laid over any flat surface. This system contains a grid of pressure sensors (48 sensors by 288 sensors) placed every 1.27 cm, producing an active measuring area of 61cm wide by 366cm long. The sampling rate is 30 Hz. Subjects were instructed to walk along the mat at their normal, self-selected, walking speed. Subjects did not wear shoes so as to eliminate shoe stiffness effects.[27] Six passes along the electronic mat were executed for each subject. For each pass, approximately eight total left and right footfalls were collected for each subject, resulting in approximately 48 footfall data points per subject. For each subject, the average and standard deviation of each gait parameter were computed.

The variables of interest for this particular study were gait speed, stride length, stride width, swing time, stance time and double support time, and the variability of these same gait parameters. These variables were selected because they have been widely studied in the literature. Gait speed is the average velocity over the distance traveled in each pass (3.66 meters). Stride length is measured on the line of progression between the heel points of two consecutive footfalls of the same foot. Stride width is the side-to-side distance between the heel of the current foot and heel of the next opposite foot. Double support time is the time elapsed between first contact of the current footfall and last contact of the previous footfall, added to the time elapsed between the last contact of the current footfall and the first contact of the next footfall. Swing time is the time a foot is not in contact with the ground. Stance time is the time duration a given footfall is in contact with the ground. The standard deviation, calculated using all of the valid steps collected for each individual, was used to reflect variability for each of the gait parameters.[28]

Fall Incidence. Falls surveillance had been ongoing for the year prior to gait testing and was continued in a similar manner for the year following gait testing. Subjects received diaries in which to record details of a fall as soon as it occurred. Once every two months, a stamped postcard was mailed to participants asking if they had fallen during that two-month period. If a "yes" response was returned, subjects were phoned to learn the details of the fall/s (e.g. cause, direction, injuries). Eighty-five percent of all cards were returned within three weeks. Those participants not returning a card were called to determine their fall status for that two-month period.

Fall direction was ascertained by an open-ended question asked of the faller or a reliable witness: What direction did you fall? Responses were coded as forward, backward, sideways, straight down, other or unknown. Anyone experiencing at least one fall in the sideways direction was classified as a side-faller. If an individual experienced only one fall over the observation period *and* indicated the direction of that fall was unknown, fall direction classification could not reliably be determined and they were not included in the analysis.

DATA ANALYSIS

All analyses were performed using SPSS Release 12.0 for Windows (SPSS Inc., Chicago, IL). All data were explored for normality. Variables with positively skewed distributions were log10 transformed prior to inferential analyses. Relationships between variables were determined using linear regression. Differences in age, weight, height and leg length between side-fallers and other-directed fallers were assessed using independent t-tests. Differences in gait characteristics between side-fallers and other-directed fallers were assessed using ANCOVA controlling for leg length, gait speed, and age where appropriate. Covariates were determined using regression analysis and graphical analyses were carried out to confirm linearity of the covariates to outcome variables of interest. The extent to which gait variables provide independent information regarding fall direction was characterized using discriminant analysis. A P-value of < 0.05 was required for analyses to be considered statistically significant.

RESULTS

Over the two-year surveillance period, 29 subjects fell to the side and 64 subjects fell in "other" directions (forward, backward, straight down). The remaining 46 subjects experienced no falls. Subsequent analyses were conducted only on the 93 subjects experiencing a fall.

There were no differences in age, weight, height or leg length between side-fallers or other-directed fallers (Table 1.1). As expected, age was significantly correlated with a number of gait variables (Table 1.2). With age, stride length decreases and gait speed slows. The amount of time spent in single leg stance and in double support also increases with age, as does the variability of time spent in stance and the variability of swing time. Additional analyses including these variables were conducted using age as a covariate. No other gait variables showed a significant association with age among this sample of fallers.

Table 1.1 Characteristics of fallers stratifying by fall direction

	Side-Fallers (N=29)	Other-Direction Fallers (N=64)	P-value[a]
Age(years)	82.3(5.1)	83.9 (5.8)	0.19
Weight (kg)	67.4 (11.1)	69.1 (12.2)	0.52
Height (cm)	161. 4 (7.7)	161.4 (9.7)	0.99
Leg length (cm)	77.4 (3.6)	76.1 (5.9)	0.20

Data are presented as mean values. Standard deviations are included in parentheses.

a P-values are for independent t-tests comparing continuous data between side-fallers and other-direction fallers who reported $\geq$ 1 fall over the two-year surveillance period.

Table 1.2 Associations between age and gait variables for all fallers

	Age (years)	P-value [a]
Stride-to-stride average		
Stride length (cm)	-.515	<0.001
Stride width (cm)	.196	0.06
Swing time (sec)	-.028	0.788
Stance time (sec)	.344	0.001
Double support time (sec)	.397	<0.001
Gait speed (m/sec)	-.433	<0.001
Stride-to-stride standard deviation		
Stride length (cm)	.078	0.064
Stride width (cm)	.061	0.562
Swing time (sec)	.201	0.05
Stance time (sec)	.351	0.001
Double support time (sec)	.093	0.374
Gait speed (m/sec)	.037	0.726

[a] P-values are for regression analyses indicating the strength and nature of the relationship between age and gait variables for all fallers

Body height was not related to gait parameters in our dataset, however leg length was associated with stride length, width, gait speed, swing time, and double support time ($p<0.05$). Thus, characteristics of gait were analyzed using leg length as a covariate for statistical analyses.

Analysis of covariance (controlling for age and leg length) revealed no differences in gait speed between the groups of fallers ($p=0.898$). The average walking speeds were 1.01 $\pm$ 0.29 m/s and 0.99 $\pm$ 0.27 m/s among the side-fallers and the non-fallers, respectively. Gait speed was significantly associated with spatial and temporal characteristics of gait (Table 1.3) and was therefore included as a covariate in between-group analyses of gait parameters. Analysis of covariance (controlling for age, leg length and gait speed) revealed the only gait characteristic that differed between side-fallers and other direction fallers was stride width (Table 1.4). Side-fallers exhibited narrower steps compared to other-direction fallers (15.7 $\pm$ 3.1 cm vs. 18.2 $\pm$ 5.1 cm; $p=0.036$). There was no difference in the variability of stride width between these two groups.

Table 1.3 Associations between gait speed and gait characteristics for all fallers

	Gait speed (m/sec)	P-value[a]
Stride-to-stride average		
Stride length (cm)	.926	<0.001
Stride width (cm)	-.375	<0.001
Swing time (sec)	-.137	0.190
Stance time (sec)	-.771	<0.001
Double support time (sec)	-.840	<0.001
Stride-to-stride standard deviation		
Stride length (cm)	.0155	0.137
Stride width (cm)	.214	0.040
Swing time (sec)	-.267	0.010
Stance time (sec)	-.474	<0.001
Double support time (sec)	-.007	0.950

[a] P-values are for regression analyses indicating the strength and nature of the relationship between gait speed and spatial and temporal gait variables (other than gait speed) for all fallers.

Table 1.4 Gait characteristics stratifying by fall direction

	Side-Fallers (N=29)	Other-Direction Fallers (N=64)	P-value[a]
Stride-to-stride average			
Stride length (cm)	136.9 (28.9)	134.2 (27.9)	0.744
Stride width (cm)	15.7 (3.20)	18.1 (5.10)	0.035
Swing time (sec)	0.433 (0.03)	0.42 (0.04)	0.179
Stance time (sec)	0.65 (0.11)	0.65 (0.11)	0.778
Double support time (sec)	0.32 (0.08)	0.33 (0.09)	0.423
Gait speed (m/sec) [b]	0.99 (0.27)	1.01 (0.29)	0.898
Stride-to-stride standard deviation			
Stride length (cm)	6.4 (3.50)	5.7 (2.90)	0.361
Stride width (cm)	2.4 (0.89)	2.3 (0.95)	0.743
Swing time (sec)	0.04 (0.03)	0.04 (0.02)	0.314
Stance time (sec)	0.05 (0.03)	0.05 (0.03)	0.720
Double support time (sec)	0.06 (0.04)	0.06 (0.04)	0.869
Gait speed (m/sec) [c]	8.09 (3.70)	7.96 (3.69)	0.785

[a] P-values are for tests of analysis of covariance comparing continuous data between side-fallers and other-direction fallers who reported $\geq$ 1 fall over the two-year surveillance period. Covariates included age, leg length and gait speed.

[b] Analysis of covariance examining differences between side-fallers and other-direction fallers in gait speed and variability of gait speed included age and leg length as covariates.

[c] Analysis of covariance examining differences between side-fallers and other-direction fallers in gait speed and variability of gait speed included age and leg length as covariates.

Group membership (i.e., side-faller or other-directed faller) was predicted by stepwise discriminant analysis using only stride width ($p=0.020$, Wilk's lambda = 0.942) such that 20 of 29 side-fallers (69%) and 41 of 64 other-directed fallers (65%) were properly classified based on stride width. Forcing additional variables such as age, gait speed and stride length into the discriminant function did not improve the extent to which the side-fallers and other-direction fallers could be correctly classified by discriminant analysis.

DISCUSSION

The results suggest that individuals with a tendency to lose balance in the sideways direction exhibit different gait characteristics in the medio-lateral direction compared to individuals who do not experience falls to the side. Specifically, side fallers walk with narrower steps compared to individuals who fall forward, backward, or straight down. Furthermore, stride width appears to be a sensitive descriptor of fall status and may be useful in identifying individuals vulnerable to high-risk falls to the side.

Increased variability of both temporal and spatial gait characteristics have been consistently associated with fall risk.[14, 28, 29, 30] The association however, has not been differentiated relative to the direction of a fall. We found that although gait variability increased with age, there were no associations between measures of variability and fall direction in our sample of fallers. Though we expected stride width variability would be greater among side-fallers, we did not observe this in our sample. However, the values for stride width variability among all fallers in our study (2.35 $\pm$ 0.93 cm) were similar to those reported by Maki (2.2 $\pm$ 0.9 cm) and Owings (2.25 $\pm$ 0.64 cm).[28, 30] This suggests that while fallers in general may exhibit greater variability in stride width compared to non-fallers, variability in stride width does not differ between individuals who fall sideways compared to those who fall in other directions.

While there have been no reports suggesting stride width is related to fall direction, one other study has reported that stride width is predictive of falls incidence.[30] Researchers reported that fallers walked with wider strides compared to non-fallers and that wider strides were not reflective of increased stability. In our study, stride widths among fallers were comparable to those reported previously among fallers: 15.7 $\pm$ 3.1 cm and 18.2 $\pm$ 5.1 cm for side-fallers and other-direction fallers, respectively; 17.4 $\pm$ 4.1 for all fallers in our study, compared to 16.0 $\pm$ 4.0

cm reported for fearful fallers.[30] However when we stratified fallers by fall direction, we found those exhibiting narrower strides had a tendency to fall sideways. It is possible that a wider gait pattern increases medio-lateral stability, explaining why individuals who exhibit a wider gait pattern did not experience high-risk falls to the side. It has been hypothesized that fallers adopt a wider stride to control the laterally displacing center of mass (COM) during the single-leg support phase of gait and that increasing step width will promote stability as a possible compensation for instability.[30, 31] Continuing along this vein, it is possible that side-fallers, who walked with narrower strides compared to other-direction-fallers in our study, are not compensating for instability in the medio-lateral direction by increasing stride width. As a result, the COM may fall closer to the outside edge of the supporting foot during the single-support phase of gait, increasing the likelihood of a fall in the lateral direction. Some evidence in support of this was highlighted in a pilot study examining the distance the COM is from the outside edge of the supporting foot while walking (a distance we call the ML safety margin) in a sample of three young and three older adults.[32] Though there were no differences between young and old subjects, the older side-faller exhibited a narrower medio-lateral safety margin (73.8 mm) compared to the older non-faller (80.1 mm) and the older other-direction faller (85.4 mm). There were too few subjects to draw any conclusions, but in light of the present study results, it may be interesting to investigate the relationship between stride width and lateral control of the center of mass in a population of older fallers.

There are a number of factors to consider when interpreting the results of this study. A potential limitation pertains to the manner in which gait parameters were obtained. Gait characteristics were determined from approximately 48 steps per subject using a gait mat system. Others have reported data based on a few (less than twelve) to hundreds, if not thousands of steps, rendering direct comparisons difficult.[14, 29, 30, 32, 31] A study by Owings and colleagues found that accurate measures of step kinematic variability required at least 400 steps when walking on an instrumented treadmill.[35] Whether a similar number of steps is required during overground walking is, as of yet, undetermined. Furthermore, methodologies differ (e.g. gait mats, instrumented treadmills, pressure sensitive insoles) and have yet to be directly compared. As such, it is unknown whether similar information regarding gait characteristics is obtained using various methods. However, our methodology was similar to the research by Maki showing that stride width was related to falls incidence and that of Brach et al. showing that stride width variability was associated with falls.[30, 33]

Another consideration is the manner in which fall status is determined. We were able to categorize falls status based on both historical and prospective data since we recruited our subjects from an ongoing longitudinal fall surveillance study. Classification of fall status was based on short-term recall as subjects were encouraged to log all falls when they occurred, and were then questioned in detail every two months about falls they reported. Thus we feel confident that individuals were accurately classified relative to fall direction. Even so, we had to include individuals reporting a single fall over the two-year period in order to have sufficient power to examine gait characteristics relative to fall direction. Ideally, we would have preferred to include only those individuals with multiple side-falls compared to individuals with multiple other-direction falls, however only 16 side-fallers experienced multiple side-falls compared to 35 other-direction-fallers with multiple falls. When examining gait characteristics between these groups who experienced multiple falls, the results were similar to those reported; however, due to the limited power of the analyses they are not presented in this paper.

Finally, the participants of this study had to be part of an ongoing fall surveillance study to participate, had to be able to walk 4 meters without the use of an assistive device, and were in continual contact with the researchers over the course of two-years. These are all factors that potentially bias the subject pool and must be considered when generalizing the study results.

The findings from this study indicate stride width is associated with side-fall incidence among older fallers walking at a “normal”, self-selected walking speed. Specifically, side-fallers, who walked with narrower strides compared to those who fell in other directions (forward, backward, straight down), may not be adapting their gait to compensate for deficits in medio-lateral stability. Further research is needed to elucidate relationships between stride width and the lateral position of the center of mass in side-fallers to determine if a narrow gait contributes to unstable walking patterns.

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THE APPLICATION OF NON-LINEAR FEEDBACK CONTROLTO GAIT ANALYSIS

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ABSTRACT

One of the fundamental tools for human gait analysis is dynamic simulation. Using advanced three dimensional motion capture systems, position and orientation of body systems can be dynamically measured with high fidelity. Traditional gait analysis incorporates motion capture measurements with inverse rigid body dynamics to determine forces and moments throughout the analysed body during specific motions such as walking, running, jumping, etc. The work reported here takes a different approach to the same problem where motion capture data is combined with a forward dynamic simulation driven by a non linear motion control system. The non-linear control system determines composite muscle forces and moments so that the simulated time history matches or tracks the measured motion captured data. Angles and moments which were obtained from the new forward dynamics method were compared with the angles and moments which were measured and calculated from inverse dynamics method for verification purpose.

NOMENCLATURE

$\vec{a}_{ci}(=a_{cix}\bar{i}_I+a_{ciy}\bar{j}_I+a_{cix}\bar{k}_I)$	3D vector of acceleration from the mass center of the *i*th link
$\vec{\alpha}_i(=\alpha_{ix}\bar{i}_{Bi}+\alpha_{iy}\bar{j}_{Bi}+\alpha_{iz}\bar{k}_{Bi})$	3D vector of angular acceleration for the *i*th link
$\vec{F}_i(=F_{ix}\bar{i}_I+F_{iy}\bar{j}_I+F_{ix}\bar{k}_I)$	3D vector of applied force on the *i*th link
$\vec{H}_{B_i/I}\oplus_i(=h_{xi}\bar{i}_{Bi}+h_{yi}\bar{j}_{Bi}+h_{zi}\bar{k}_{Bi})$	3D vector of angular momentum for the *i*th link
$\vec{\omega}_i(=p\bar{i}_{Bi}+q\bar{j}_{Bi}+r\bar{k}_{Bi})$	3D vector of angular velocity for the *i*th link
$\dfrac{{}^{I}d\vec{H}_{B_i/I}\oplus_i}{dt}=\dfrac{{}^{B}d\vec{H}_{B_i/I}\oplus_i}{dt}+\vec{\omega}_{Bi/I}\times\vec{H}_{Bi/I}\oplus_i=\tilde{h}_{xi}\bar{i}_{Bi}+\tilde{h}_{yi}\bar{j}_{Bi}+\tilde{h}_{zi}\bar{k}_{Bi}$	3D vector of angular momentum derivative of the *i*th link with respect to the global frame $\begin{Bmatrix}\tilde{h}_{xi}\\ \tilde{h}_{yi}\\ \tilde{h}_{zi}\end{Bmatrix}=\mathrm{I}_i\begin{Bmatrix}\dot{p}\\ \dot{q}\\ \dot{r}\end{Bmatrix}+S_{\omega i}\mathrm{I}_i\begin{Bmatrix}p\\ q\\ r\end{Bmatrix}$ $S_{\omega i}=\begin{bmatrix}0&-r&q\\ r&0&-p\\ -q&p&0\end{bmatrix}$
$\vec{r}_{ci\to j}$	3D vector from the mass center of the *i*th segment to the *j*th joint point in the global frame
$\vec{r}_{ci\to j(B)}$	3D vector from the mass center of *i*th segment to the *j*th joint point in the body frame
$\vec{R}_i(=R_{ix}\bar{i}_I+R_{iy}\bar{j}_I+R_{ix}\bar{k}_I)$	3D vector of reaction force on the *i*th link at the *i*th joint
$\vec{T}_i$	3D vector of applied torque on the *i*th link at the *i*th joint
$\vec{w}_i(=w_i\bar{k}_I)$	The vector of weight for the *i*th link
C_A	Matrix for the accelerations relating kinematic

	relation
C_T	Matrix for the reactions relating translational dynamics
I_i	The matrix for mass moment of inertia for the *i*th link
M	The matrix for segments mass
T_S	The distances matrix between mass centers and joint points relating angular acceleration
S_T	The matrix of relating centripetal acceleration
Ro_i^T	Inverse rotation matrix for *i*th segment
B	The matrix of relating joint torques
S_ω	The matrix of relating the angular velocities
S_R	The matrix of relating the reaction forces
S_f	The matrix of relating the applied forces

INTRODUCTION

Characterizing the gait pattern is the goal of many gait researchers. Joint moments obtained from gait analyses are commonly used to characterize normal and pathologic gaits and to make comparisons within and between populations.[1] The muscle moments of force provide a window on the net muscular activity about a joint.[2] Mechanical power can be calculated from the obtained joint moment with angular velocity, and the mechanical energy consumption during human gait can be estimated. Proper and efficient calculation of the required joint moment of lower extremities during gait is the most basic and important task for common gait analysis.

Inverse dynamics method is widely used for moment calculation because of its simplicity and ease in application and had been chosen by many previous researchers.[3, 4, 5, 6] Inverse dynamics is a technique of computing temporal patterns of joint moments from observed motions of body segments. Linear and angular accelerations of each body segment and reaction forces from foot contact to ground are required for inverse dynamics method. For instance, kinematics information of the foot, combined with the reaction forces from the ground, determines the necessary moments and forces from the ankle. The kinetic information obtained from the ankle is used for kinetic calculations of the shank. These same procedures repeat from shank to thigh, and from thigh to hip.

Forward dynamics simulates motions by adjusting the kinetics of segments until those motions are accurately following the motions of human movement, which is in concern. The kinetic and kinematic equations, which are representing a dynamic system, are obtained. The motion of the concerned system can be simulated by numerical integration from the obtained equations of motion.

The inverse dynamics method provides an easier way to attain the required kinetics without deriving complicated equations of motion, which is necessary for forward dynamics method. Double numerical differentiations for all positions and rotations of body segments are required for the inverse dynamics method. Error propagation is a fatal drawback of the inverse dynamics method. Errors of the distal joint calculations affect those at more proximal calculations. Forward dynamics is potentially a very powerful and accurate method without error propagation. More detailed analysis is possible with the forward dynamics method. Complete mathematical expressions for the human body and predicting appropriate input are essential for the forward dynamics method. Different types of forward dynamics methods were used in previous research depending on the purposes of the studies. Many of them developed the model in simplified patterns (analysis in sagittal plane only, with a small number of segments), and the required kinetics (moments or forces) for the simulation of motion were taken from inverse dynamics method because predicting all proper inputs for the required movements was not easy.[7-9] Increasing numbers of recent studies are using muscle-driven forward dynamic simulations because it affords the researchers access to many mechanical variables of interests.[10-12] The over defined problem of muscle activities and computational expense are the tasks to be solved for this method. A study by Russell used simulation of forward dynamic model with non-linear feedback control.[13] Simplicity of the model of Russell's simulation makes the kinetic information of locomotion not suitable for the real human gait analysis.

The new method of joint moment calculation is introduced in this study without error propagation problems and without difficulty of predicting proper inputs. Forward dynamics method combined with linearized feedback control system (FDMLFC) was applied in this study for mechanical joint moment calculation. Feedback linearization amounts to canceling the nonlinearities in a non-linear system so that the closed-loop dynamics is in a linear form.[14] The humanoid dynamics model was developed with twelve connected rigid links in three-dimensional space. Simulation with non-linear feedback control was performed, and required moments from all joints were obtained as a result of control efforts. FDMLFC provides improved understandings of human movements as a complete function of relations between kinetics and kinematics.

METHOD

DYNAMIC MODEL

Twelve rigid body segments were chosen to represent a humanoid dynamic model. For simplicity, head and hands were assumed to be rigidly attached to the upper torso and forearms, respectively (Figure 2.1). Rotations in three directions were allowed by using gimbal joints between the segments.

Before deriving the whole humanoid model, a simpler model with two rigid body segments was considered to demonstrate the method. The degree of freedom (DOF) for the two rigid link system is twelve (6 from rotation and 6 from translation) in three dimensional space. Assuming the fixed upper end point and chained connection between segments, the DOF was reduced by 3 from each segment, and the DOF of the system changed to six (3D rotations of two segments). The kinetic and free body diagram for each segment of the system was first considered as shown in Figure 2.2. The body frame of each segment was initially aligned with the global frame. By applying Newton's second law, the translational dynamics for two segments were obtained and shown in equation (1a), and (1b).

$$-\vec{R}_1 + \vec{R}_2 + \vec{F}_1 + \vec{w}_1 = m_1\vec{a}_{c1} \tag{1a}$$

$$\vec{R}_2 + \vec{F}_2 + \vec{w}_2 = m_2\vec{a}_{c2} \tag{1b}$$

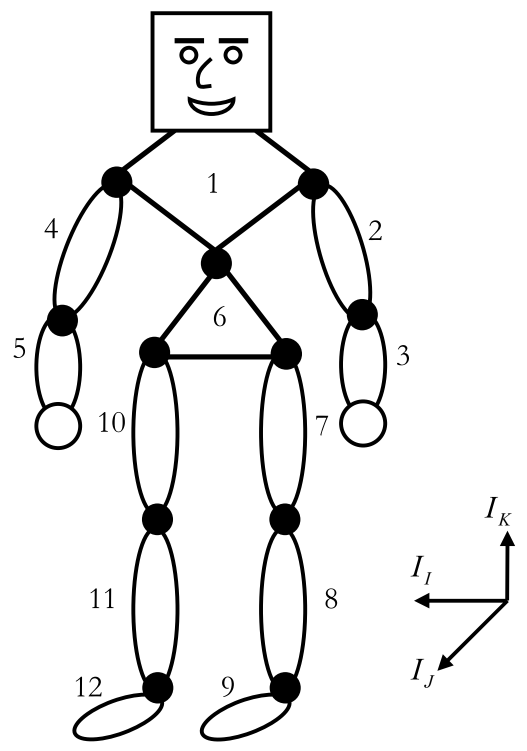


Figure 2.1 Humanoid dynamic model with twelve rigid segments

The rotational dynamics for the system was achieved by equation (2a), and (2b) by summing up moments about each mass center of segment. $\vec{r}_{ci \to j}$ (which is the vector from the *i-th* segment mass center to the *j-th* joint) was expressed with respect to (WRT) the global frame by applying rotational transformation calculated from relative angles of the body frame WRT the global frame.

$$-\vec{r}_{c1\to1} \times \vec{R}_1 + \vec{r}_{c1\to2} \times \vec{R}_2 + \vec{r}_{c1\to1*} \times \vec{F}_1 - \vec{T}_1 + \vec{T}_2 = \frac{{}^{I}d\vec{H}_{B_1/I}\ \oplus_1}{dt} \tag{2a}$$

$$-\vec{r}_{c2\to1} \times \vec{R}_2 + \vec{r}_{c2\to2*} \times \vec{F}_2 = \frac{{}^{I}d\vec{H}_{B_2/I}\ \oplus_2}{dt} \tag{2b}$$

Kinematic relations between segments which are shown in equations (3a), (3b), and (3c) provide constraints which connect each segment as connected chain.

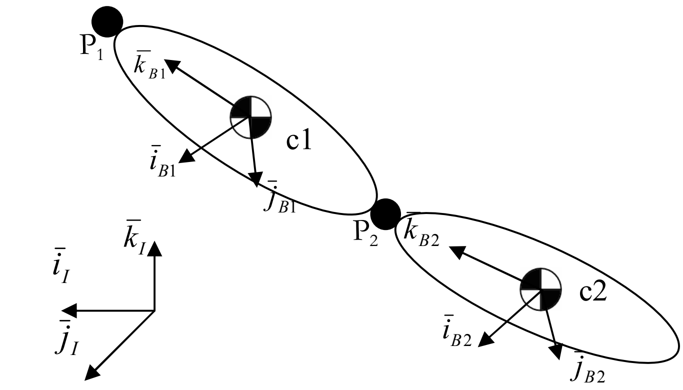
$$\vec{a}_1 = \vec{a}_{c1} + \vec{\alpha}_1 \times \vec{r}_{c1\to1(B)} + \vec{\omega}_1 \times (\vec{\omega}_1 \times \vec{r}_{c1\to1(B)}) \tag{3a}$$

$$\vec{a}_2 = \vec{a}_{c1} + \vec{\alpha}_1 \times \vec{r}_{c1\to2(B)} + \vec{\omega}_1 \times (\vec{\omega}_1 \times \vec{r}_{c1\to2(B)}) \tag{3b}$$

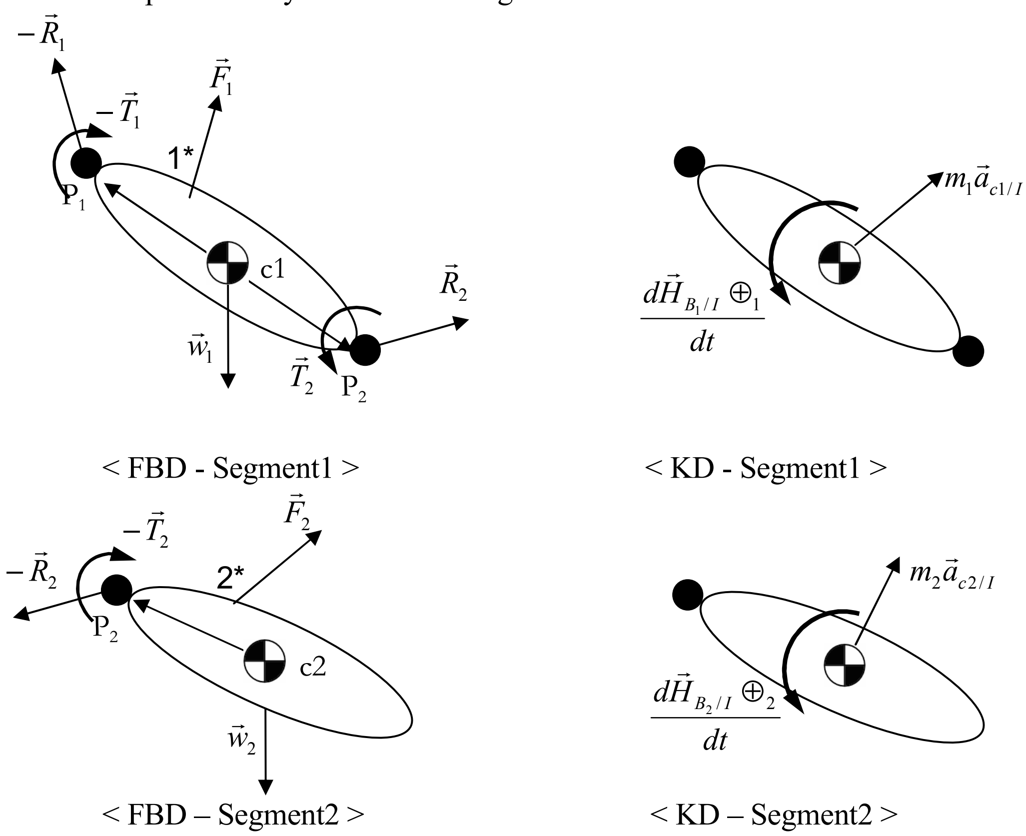
$$\vec{a}_2 = \vec{a}_{c2} + \vec{\alpha}_2 \times \vec{r}_{c2\to2(B)} + \vec{\omega}_2 \times (\vec{\omega}_2 \times \vec{r}_{c2\to2(B)}) \tag{3c}$$

The common acceleration $\vec{a}_2$ from equation (3b) and equation (3c) was eliminated and a new equation representing (3b) and (3c) was obtained as shown in equation (3d).

$$\vec{a}_{c2} - \vec{a}_{c1} + \vec{\alpha}_2 \times \vec{r}_{c2\to2(B)} - \vec{\alpha}_1 \times \vec{r}_{c1\to2(B)} + \vec{\omega}_2 \times (\vec{\omega}_2 \times \vec{r}_{c2\to2(B)}) - \vec{\omega}_1 \times (\vec{\omega}_1 \times \vec{r}_{c1\to2(B)}) = 0 \tag{3d}$$



a. Open chain system with two rigid links



b. Individual segment free body diagrams and kinetic diagrams

Figure 2.2 Simple open chain system with two rigid links

One set of kinematic equations and two sets of kinetic equations from translational dynamics and rotational dynamics constitute all of the equations of motion for the two connected rigid link system. By introducing matrix formation, obtained equations of motion can be expressed as below

$$C_T\overline{R} = M\overline{A} - \overline{F} - \overline{W} \tag{4a}$$

$$Ro^T I\dot{\overline{\omega}} + Ro^T S_\omega I\overline{\omega} = S_R\overline{R} + S_F\overline{F} + B\overline{T} \tag{4b}$$

$$C_A\overline{A} + T_S\dot{\overline{\omega}} + \overline{S}_T = \overline{a} \tag{4c}$$

The acceleration vector ($\overline{A}$) from equation (4c) was substituted into equation (4a), and the obtained reaction vector ($\overline{R}$) from equation (4a) was substituted into equation (4b), and a dynamic equation of motion in matrix formation was obtained as shown below.

$$\begin{aligned} &\left(Ro^T I + S_R C_T^{-1} M C_A^{-1} T_S\right)\overline{\Omega} = \\ &\qquad S_R C_T^{-1} M C_A^{-1}\left(\overline{a} - \overline{S}_T\right) - S_R C_T^{-1}\left(\overline{F} + \overline{W}\right) + S_F\overline{F} + B\overline{T} - Ro^T S_\omega I\overline{\omega} \end{aligned} \tag{5}$$

Components of matrix and column vector of equation (5) are shown in Appendix A. Equation (5) is a first order differential equation of angular velocity ($\Omega = \dot{\omega}$). The same method was applied to the humanoid model consisting of twelve rigid links with the following differences: expanded DOF, modified matrices for kinetic and kinematic equations, and applied inputs from the foot. Matrices for the humanoid model are shown in Appendix B.

LINEARIZED FEEDBACK CONTROL

Linearized feedback control law is applied to the non linear system by converting the non linear system into a related linear system. Feedback linearization amounts to canceling the nonlinearities in a nonlinear system so that the closed-loop dynamics is in a linear form.[14]

A brief idea of the feedback linearization method is shown in Figure 2.3. The nonlinear part of the equation of motion is converted into pseudo control (γ) as shown in equation (6a). Second order linear feedback control law is designed with two gains to track desired angles and angular velocities (equation (6b)). Physical control efforts are obtained by converting pseudo control to non linear domain (equation (6c)).

$$\ddot{\theta} = a^{-1}(b + BT) \rightarrow \ddot{\theta} = \gamma \tag{6a}$$

$$\gamma = \ddot{\theta}_D - k_0 \cdot (\theta - \theta_D) - k_1 \cdot (\dot{\theta} - \dot{\theta}_D) \tag{6b}$$

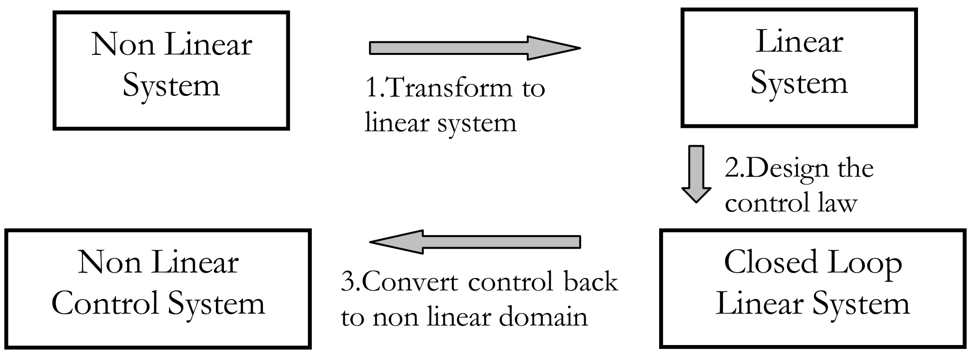
$$BT = a\gamma - b \tag{6c}$$

where, θ_d : Desired angle. γ : Pseudo control

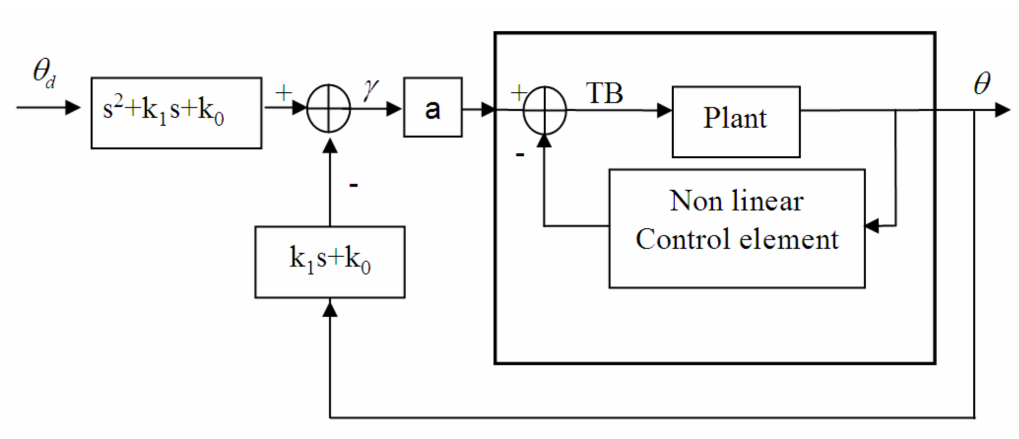
$$\underbrace{(Ro^T I + S_R C_T^{-1} M C_A^{-1} T_S)}_{a}\dot{\omega} =$$

$$\underbrace{S_R C_T^{-1} M C_A^{-1}\left(\overline{a} - \overline{S}_T\right) - S_R C_T^{-1}\left(\overline{F} + \overline{W}\right) + S_F \overline{F} - Ro^T S_\omega I \overline{\omega} - \overline{T}_d}_{b} + B\overline{T}$$

The control gains k_0, and k_1 were selected from pole placement to stabilize the tracking error dynamics and to achieve suitable performance (k_0 = 4224, k_1 = 130).



a. Basic idea for feedback linearization method



b. Block diagram of non linear feedback linearization

Figure 2.3 Linearized feedback control system

EXPERIMENTAL MEASUREMENTS

Developed humanoid dynamics model and non linear control system require the desired angles for tracking to simulate real human gait. For the collection of required motion data, 37 markers were attached to the participant, who was then asked to perform a normal walk of approximately 4 meters on a straight and level path. The 3D motion capture system (Vicon Motion System, Oxford, United Kingdom) recorded the positions of all markers and two force plates mounted on the middle of the path calibrated ground reaction forces synchronized with motion capture data. The sampling frequency for each camera was 60Hz, and 600Hz for each force plate. Hip, knee, and ankle joint center points were computed from the positions of markers attached to the participant's body (Appendix C). The body frame, fixed on rigid body segment, was obtained from three points attached in one plane of each segment. The relative rotations of body frames with respect to the global frame provide the angle of segments on the body. Moments from foot contact to the ground were calculated based on measured reaction forces and the center of pressure position from force plates.

SIMULATION BY NUMERICAL INTEGRATION

A fourth-order Runge-Kutta formula was applied for numerical integration of the obtained dynamic model. Measured orientations for all segments were set as desired properties for feedback control to track. Instead of Euler angles, equivalent quaternion was used for integration of the system to avoid the singularity which might happen with an angle of 90 degrees in the middle (or second) rotation of Euler angles. The relationship between angular velocity and quaternion is shown below in equations (8a) and (8b). By substituting angles and angular velocities in quaternion into the original equation of motion (equation (5)), the new equation of motion in quaternion was obtained and used for numerical simulation.

$$\begin{bmatrix} 0 \\ p \\ q \\ r \end{bmatrix} = 2 \cdot \begin{bmatrix} q_0 & q_1 & q_2 & q_3 \\ -q_1 & q_0 & q_3 & -q_2 \\ -q_2 & -q_3 & q_0 & q_1 \\ -q_3 & q_2 & -q_1 & q_0 \end{bmatrix} \cdot \begin{bmatrix} \dot{q}_0 \\ \dot{q}_1 \\ \dot{q}_2 \\ \dot{q}_3 \end{bmatrix} \tag{8a}$$

$$\begin{bmatrix} 0 \\ \dot{p} \\ \dot{q} \\ \dot{r} \end{bmatrix} = 2 \cdot \begin{bmatrix} q_0 & q_1 & q_2 & q_3 \\ -q_1 & q_0 & q_3 & -q_2 \\ -q_2 & -q_3 & q_0 & q_1 \\ -q_3 & q_2 & -q_1 & q_0 \end{bmatrix} \cdot \begin{bmatrix} \ddot{q}_0 \\ \ddot{q}_1 \\ \ddot{q}_2 \\ \ddot{q}_3 \end{bmatrix} + 2 \cdot \begin{bmatrix} \dot{q}_0 & \dot{q}_1 & \dot{q}_2 & \dot{q}_3 \\ -\dot{q}_1 & \dot{q}_0 & \dot{q}_3 & -\dot{q}_2 \\ -\dot{q}_2 & -\dot{q}_3 & \dot{q}_0 & \dot{q}_1 \\ -\dot{q}_3 & \dot{q}_2 & -\dot{q}_1 & \dot{q}_0 \end{bmatrix} \cdot \begin{bmatrix} \dot{q}_0 \\ \dot{q}_1 \\ \dot{q}_2 \\ \dot{q}_3 \end{bmatrix} \tag{8b}$$

where $\overline{\omega}_{B/I} = p\,\bar{I}_B + q\,\bar{J}_B + r\,\overline{K}_B$ (angular velocity)

Simulation of the two link model was performed first to verify the method. The physical properties (weight, length, and moment of inertia) of two links were set close to a real human leg (thigh, and shank).

The ground forces from force plate to the foot are important kinetics for the simulation of humanoid dynamic model. The forces and moments from ground reaction forces were calculated based on the force data and position data for center of pressure from two force plates. For the simulation of two complete stance walking, forces and moments before the first stance and after the second stance were needed. (During double support period) Those required forces and moments were adapted from those, which were captured from first and second stance steps from force plate, and synchronized with the time of ground contacts of each foot.

In the numerical simulation, one step in reference angles (the desired angles for tracking) was divided in fifteen steps so the tracking simulation advance done step in desired angles after 15 repeated tracking simulations. Those fourteen approximated data points inside of the one step of reference angles were obtained by linear fitting.

For verification of the method, the simulated results were compared with the results from inverse dynamics method. Same kinematic model and same convention for rotation were applied to both methods for comparison purpose. The rotation sequence for the thigh and shank was about x, y, z axes, and about x, z, y axes for the foot. The FDMLFC method calculated the moments with respect to global frame. Those moments in global frame were translated to each body frame, which the rotation of each segments were based on. The moments in sagittal plane for hip, knee, and ankle were obtained by dot products of moments in global frame and X-axes in distal segment (I_d in Figure 2.4). The moment in frontal plane for hip was calculated by dot product of hip moment in global frame and the intermediate unit vector which was obtained from the cross product of the Z-axis in distal segment and X-axis in proximal segment (K_d and I_P from Figure 2.4 respectively).[15] The moment in frontal plane for ankle was calculated by dot product of hip moment in global frame with Y-axis of proximal segment (J_p in figure 2.4).

The moments and angular velocities with respect to same axes of rotations were used for mechanical power calculation.

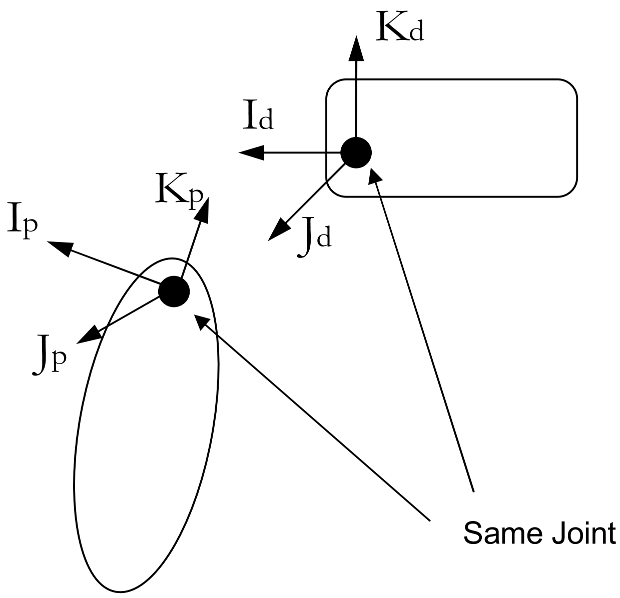


Figure 2.4 Unit vectors for coordinates of distal segment and proximal segment. (I_d, J_d, K_d are unit vectors for distal segment, and I_p, J_p, K_p are unit vectors for proximal segment)

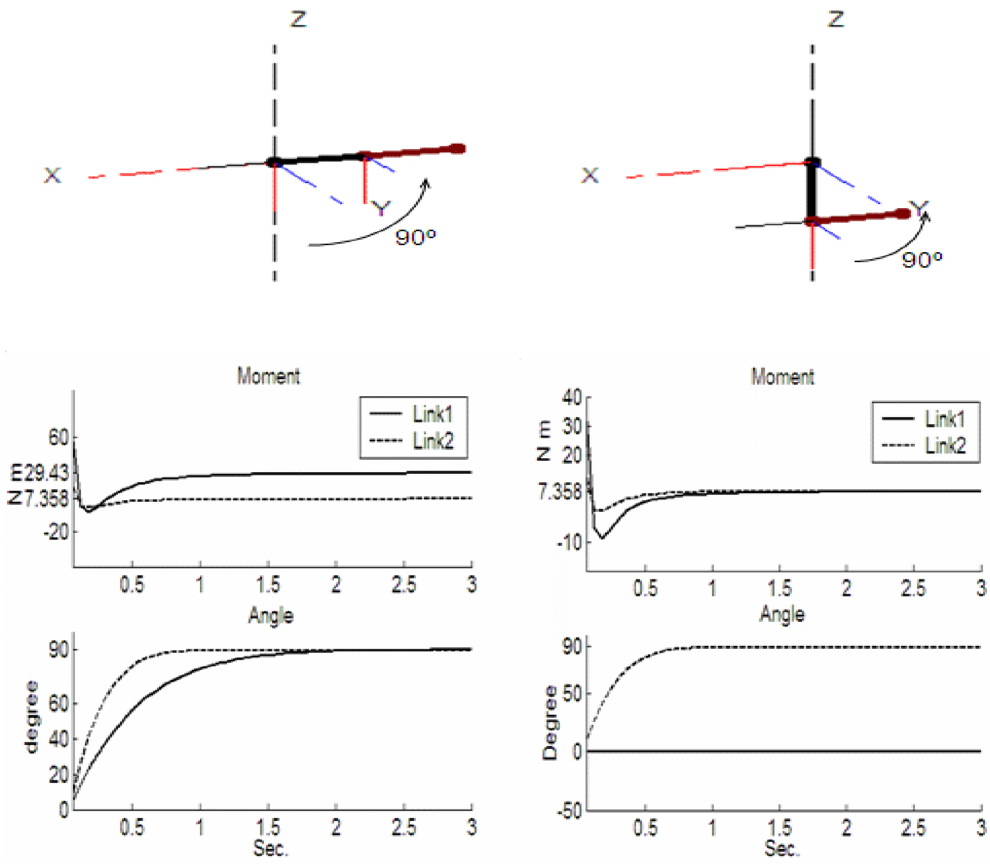
RESULTS

The simple two links model was simulated with different angles tracking. First, the static desired angles were set for the system to track as shown in Figure 2.4. The steady states tracking angles and the required moments from two joint points of links were obtained satisfactorily in both cases.

The result of simulation for the system (two links) based on the arbitrarily chosen moments (without feedback control) was compared with the simulation result of the system with feedback control to verify the simulation with feedback control for the dynamic tracking case as shown in Figure 2.5. Moments and angles in the thin line in Figure 2.5 represent the applied chosen moments for simulation and the simulation results in angles respectively. Those obtained angles as simulation results were set for the desired angles so that the system with feedback control can track. Moments and angles in the thick lines are control efforts and control results from feedback tracking control respectively. Plots of errors are the differences in moments and angles between two simulations. Errors in moments and angles are less than 2%, and those are bounded. The tracking performance was varied depending on the frequency of the desired angles. Errors were increased as the desired angles were getting faster. Simulation with feedback control also performed to track the same angles with white noise, which is in the range of possible noise from Vicon system (±0.25 degrees). The feedback tracking was not sensitive from the added noise in desired angels and showed similar error boundary with bounded errors.

The simulation of normal human gait was performed based on a 3D dynamic humanoid model and a non-linear feedback control system. Kinematic information obtained from the 3D motion capture system was used as references for the developed dynamic model to track. Numerical simulation of a single trial took about fifteen seconds. Angles from the segments of lower extremities and the required moments from the segments were shown in Figure 2.6 for sagittal plane and in Figure 2.7 for frontal plane.

Simulated results from FDMLFC (forward dynamic model combined with linearized feedback control) were compared with the results of inverse dynamics which had been obtained from the Vicon software. When compared, a similar pattern in moments illustrates that the developed humanoid dynamic model properly represents the human motion dynamics. A satisfactorily tracking simulation demonstrates that the non-linear feedback control system was properly designed.



(a) 90 degrees for both links (b) 90 degrees for link 2 only

Figure 2.4 Two link pendulum simulation (static angle tracking)

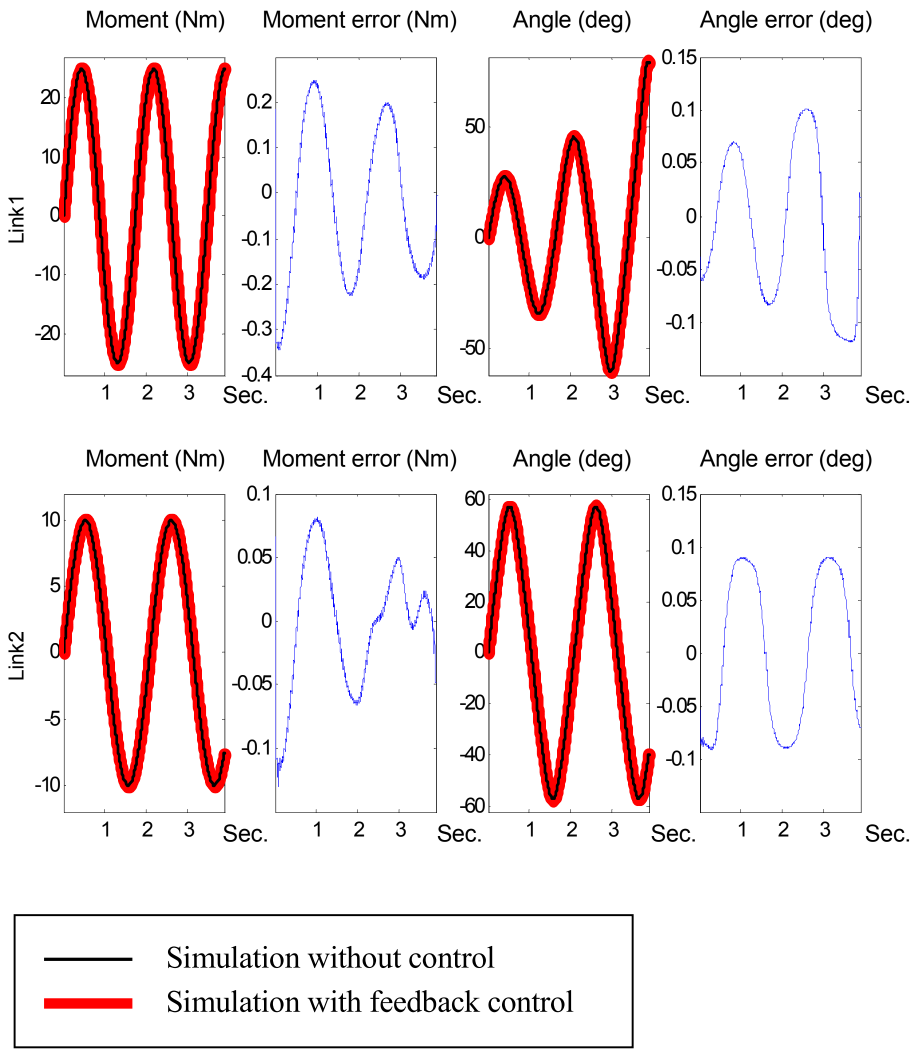


Figure 2.5 Simulation comparisons in control efforts (moments) and control results (angles)

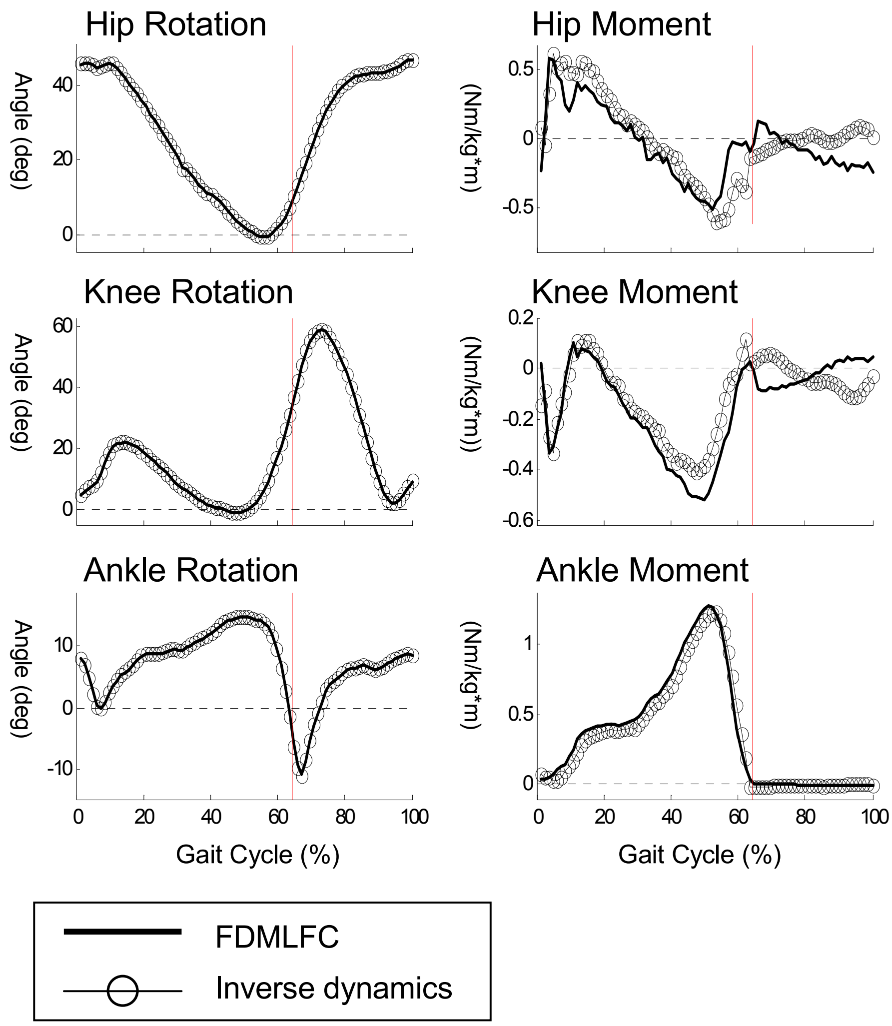


Figure 2.6 Comparisons of angles and moments in sagittal plane between FDMLFC and inverse dynamics method

(For moment plots, positive values are extensor moments, and negative values are flexor moments)

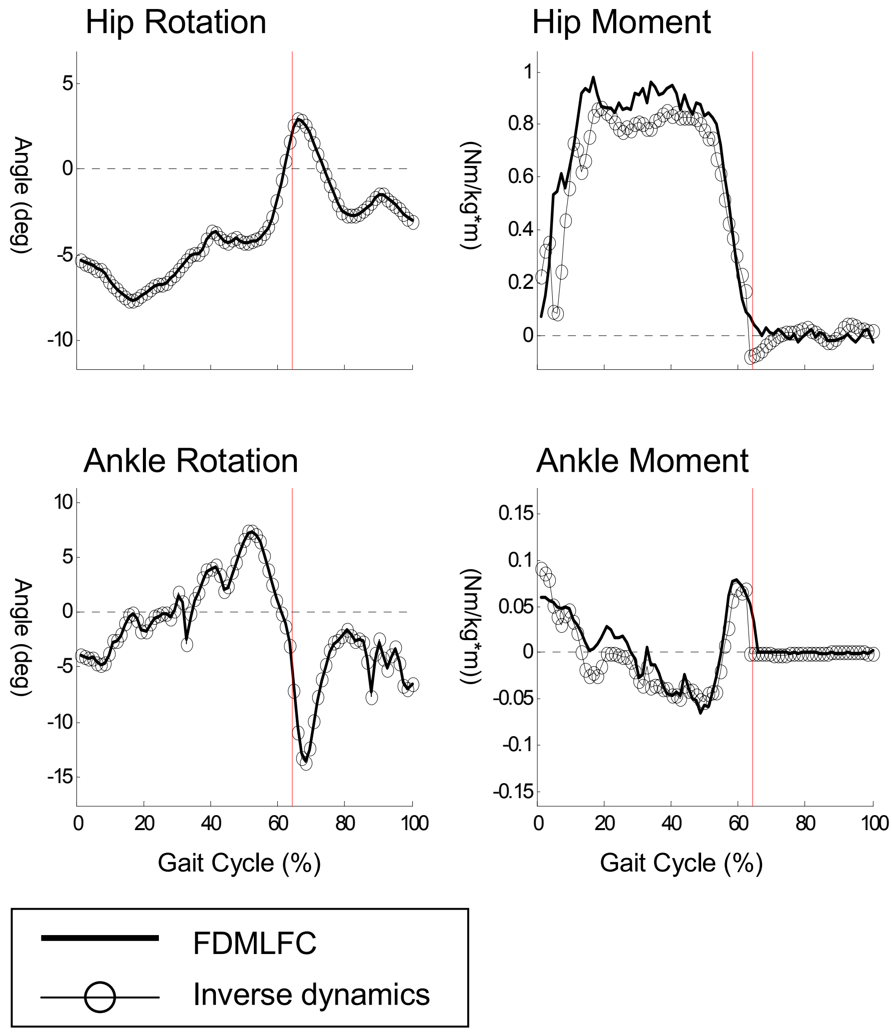


Figure 2.7 Comparisons of angles and moments in frontal plane between FDMLFC and inverse dynamics method

(For moment plots, positive value in hip is abductor moment, and positive value in ankle is evertor moment)

DISCUSSION

Calculating the required moments for the concerning joint is a basic step for analysis of human movement. The inverse dynamic method is widely used for kinetic computation because of its simplicity in understanding and application compared with the forward dynamic method. In the inverse dynamic method, required accelerations are obtained from twice numerical differentiation of measured angles. The analysis on a series of segments is generally conducted beginning with the most distal segment, proceeding proximally up to the next segment and so on.[16] The errors from the distal joint calculations affect those at more proximal calculations.

The forward dynamic method is an alternative for kinetic calculations in human movement analysis. In the forward dynamic method, we need to adjust the kinetics of segments until those motions of the segments are accurately following the motions of the human movement which is in concern. In contrast to the inverse dynamic method, the forward dynamic method uses the whole equation of motion, not separated into sequential equations. With large numbers of variables (depending on the degrees of freedom of the system), complicated mathematical expressions in differential equation form should be solved (mostly numerically). Theoretically, predicting the input required for a certain movement in a simple system is possible when the forward dynamic method is in use. However, in a multi-body system, predicting the appropriate inputs needed for all segments to follow the desired movements is almost impossible.

Predicting the required input for the simulation of forward dynamics method are the most challenging part. Many previous studies used the kinetic information which had been obtained from inverse dynamics.[7, 8, 10, 11] Initial positions of all segments for the dynamic model and those for the real human trial should be synchronized for this method to get proper results in simulation. Also achieving the desired movement is not guaranteed from those works because of the nature of feed forward control. And the required usage of the inverse dynamic method for calculating the required input could weaken the reason for using the forward dynamic method. In the study of Russeel, forward dynamics model was simulated with non-linear feedback control, but the simplicity of the model (three rigid bodies in a sagittal plane only) make the simulation not suitable for the real human gait analysis.[13]

As introduced in this study, in FDMLFC (forward dynamic method combined with linearized feedback control), control efforts were made to force the dynamic model to track the measured quaternions (obtained from measured Euler angles) of all segments individually and simultaneously, within the constraints that came from the developed equations of motion. The required gains in linearized feedback control were selected from pole placement to make the error dynamic system to be stabilized with satisfactory tracking performance. Potential singularity from gimbal lock can be avoided by using quaternions instead of Euler angles in the simulation of trials, so no limit theoretically exists for movements. This makes FDMLFC method can be used not only for the gait analysis but also any human movement analysis. Because of the applied feedback control system, the simulation of the dynamic model can be guaranteed to follow the movement in concern unlike the other forward dynamics method without feedback control.[7, 8, 10, 11] The developed 3D humanoid dynamics model with twelve segments of body in this study, provide better understanding of human movements.

Because of simultaneous tracking, FDMLFC method does not have the problem of error propagation. The FDMLFC method theoretically does not require the filtering process while the traditional inverse dynamic method requires removing accumulated noise. As the model has been expanded from a two link to a twelve link model, FDMLFC method has no limit on expanding the degree of freedom theoretically. More complicated and realistic models including the muscle driven model can be analyzed with this method.

As shown from the two link simple model simulation, the errors are varying depending on the speed of movements. Different gains or step size have to be considered depending on tests. The required input forces and moments from the contacting foot make huge limitations in the application of FDMLFC method. For the complete one cycle of gait, at least three force plates are required unless using the duplicated (or adapted) forces and moments. Also the simulation with modified parameters (like changing the mass of segments), which is based on the gait test already taken, is not possible because the changes made on ground reaction forces are not available without doing the test again.

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APPENDIX

APPENDIX A - MATRIX FOR SIMPLE SYSTEM WITH TWO CONNECTED LINKS

$$C_T = \begin{bmatrix} -1 & 0 & 0 & 1 & 0 & 0 \\ 0 & -1 & 0 & 0 & 1 & 0 \\ 0 & 0 & -1 & 0 & 0 & 1 \\ 0 & 0 & 0 & -1 & 0 & 0 \\ 0 & 0 & 0 & 0 & -1 & 0 \\ 0 & 0 & 0 & 0 & 0 & -1 \end{bmatrix} \qquad M = \begin{bmatrix} m_1 & 0 & 0 & 0 & 0 & 0 \\ 0 & m_1 & 0 & 0 & 0 & 0 \\ 0 & 0 & m_1 & 0 & 0 & 0 \\ 0 & 0 & 0 & m_2 & 0 & 0 \\ 0 & 0 & 0 & 0 & m_2 & 0 \\ 0 & 0 & 0 & 0 & 0 & m_2 \end{bmatrix}$$

$$\overline{R} = \begin{bmatrix} R_{1x} & R_{1y} & R_{1z} & R_{2x} & R_{2y} & R_{2z} \end{bmatrix}^T \quad \overline{A} = \begin{bmatrix} a_{1x} & a_{1y} & a_{1z} & a_{2x} & a_{2y} & a_{2z} \end{bmatrix}^T$$

$$\overline{F} = \begin{bmatrix} F_{1x} & F_{1y} & F_{1z} & F_{2x} & F_{2y} & F_{2z} \end{bmatrix}^T \quad \overline{W} = \begin{bmatrix} 0 & 0 & w_1 & 0 & 0 & w_2 \end{bmatrix}^T$$

$$I = \begin{bmatrix} I_{1xx} & I_{1xy} & I_{1xz} & 0 & 0 & 0 \\ I_{1xy} & I_{1yy} & I_{1yz} & 0 & 0 & 0 \\ I_{1xz} & I_{1yz} & I_{1zz} & 0 & 0 & 0 \\ 0 & 0 & 0 & I_{2xx} & I_{2xy} & I_{2xz} \\ 0 & 0 & 0 & I_{2xy} & I_{2yy} & I_{2yz} \\ 0 & 0 & 0 & I_{2xz} & I_{2yz} & I_{2zz} \end{bmatrix} \qquad S_\omega = \begin{bmatrix} & & & 0 & 0 & 0 \\ & S_{\omega 1} & & 0 & 0 & 0 \\ & & & 0 & 0 & 0 \\ 0 & 0 & 0 & & & \\ 0 & 0 & 0 & & S_{\omega 2} & \\ 0 & 0 & 0 & & & \end{bmatrix}$$

Where, $S_{\omega i} = \begin{bmatrix} 0 & -\omega_{zi} & \omega_{yi} \\ \omega_{zi} & 0 & -\omega_{xi} \\ -\omega_{yi} & \omega_{xi} & 0 \end{bmatrix}$

$$Ro^T = \begin{bmatrix} & & & 0 & 0 & 0 \\ & {Ro_1}^T & & 0 & 0 & 0 \\ & & & 0 & 0 & 0 \\ 0 & 0 & 0 & & & \\ 0 & 0 & 0 & & {Ro_2}^T & \\ 0 & 0 & 0 & & & \end{bmatrix}$$

Where, ${Ro_i}^T$: Inverse rotation matrix for *i*th segment

$$S_R = \begin{bmatrix} 0 & \Delta z_1^1 & -\Delta y_1^1 & 0 & -\Delta z_1^2 & \Delta y_1^2 \\ -\Delta z_1^1 & 0 & \Delta x_1^1 & \Delta z_1^2 & 0 & -\Delta x_1^2 \\ \Delta y_1^1 & -\Delta x_1^1 & 0 & -\Delta y_1^2 & \Delta x_1^2 & 0 \\ 0 & 0 & 0 & 0 & \Delta z_2^2 & -\Delta y_2^2 \\ 0 & 0 & 0 & -\Delta z_2^2 & 0 & \Delta x_2^2 \\ 0 & 0 & 0 & \Delta y_2^2 & -\Delta x_2^2 & 0 \end{bmatrix}$$

$\Delta x_i^j, \Delta y_i^j, \Delta z_i^j$: Distance components from mass center of the ith link to the jth joint point in the global frame

$$S_F = \begin{bmatrix} 0 & -\Delta z_1^* & \Delta y_1^* & 0 & 0 & 0 \\ \Delta z_1^* & 0 & -\Delta x_1^* & 0 & 0 & 0 \\ -\Delta y_1^* & \Delta x_1^* & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & -\Delta z_2^* & \Delta y_2^* \\ 0 & 0 & 0 & \Delta z_2^* & 0 & -\Delta x_2^* \\ 0 & 0 & 0 & -\Delta y_2^* & \Delta x_2^* & 0 \end{bmatrix}$$

$\Delta x_i^*, \Delta y_i^*, \Delta z_i^*$: Distance components from mass center to the force applied point of the ith link in the global frame

$$B = \begin{bmatrix} -1 & 0 & 0 & 1 & 0 & 0 \\ 0 & -1 & 0 & 0 & 1 & 0 \\ 0 & 0 & -1 & 0 & 0 & 1 \\ 0 & 0 & 0 & -1 & 0 & 0 \\ 0 & 0 & 0 & 0 & -1 & 0 \\ 0 & 0 & 0 & 0 & 0 & -1 \end{bmatrix} \qquad C_A = \begin{bmatrix} 1 & 0 & 0 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 & 0 & 0 \\ 0 & 0 & 1 & 0 & 0 & 0 \\ -1 & 0 & 0 & 1 & 0 & 0 \\ 0 & -1 & 0 & 0 & 1 & 0 \\ 0 & 0 & -1 & 0 & 0 & 1 \end{bmatrix}$$

$$\overline{T} = \begin{bmatrix} T_{1x} & T_{1y} & T_{1z} & T_{2x} & T_{2y} & T_{2z} \end{bmatrix}^T$$

$$T_S = \begin{bmatrix} Ro_1{}^T \begin{bmatrix} 0 & -\Delta z^1_{1(B)} & \Delta y^1_{1(B)} \\ \Delta z^1_{1(B)} & 0 & -\Delta x^2_{1\,(B)} \\ -\Delta y^1_{1(B)} & \Delta x^1_{1(B)} & 0 \end{bmatrix} & \begin{matrix} 0 & 0 & 0 \\ 0 & 0 & 0 \\ 0 & 0 & 0 \end{matrix} \\ Ro_2{}^T \begin{bmatrix} 0 & -\Delta z^2_{1\,(B)} & \Delta y^2_{1\,b} \\ \Delta z^2_{1\,(B)} & 0 & -\Delta x^2_{1\,(B)} \\ -\Delta y^2_{1\,(B)} & \Delta x^2_{1\,(B)} & 0 \end{bmatrix} & Ro_2{}^T \begin{bmatrix} 0 & -\Delta z^2_{2(B)} & \Delta y^2_{2(B)} \\ \Delta z^2_{2(B)} & 0 & -\Delta x^2_{2(B)} \\ -\Delta y^2_{2(B)} & \Delta x^2_{2(B)} & 0 \end{bmatrix} \end{bmatrix}$$

$\Delta x^j_{i\,(B)}, \Delta y^j_{i\,(B)}, \Delta z^j_{i\,(B)}$: Distance components from mass center of the ith link to the jth joint point in the body frame

$$\bar{S}_T = \begin{Bmatrix} Ro_1{}^T S_{\omega 1} S_{\omega 1} \{\bar{d}^1_1(b)\} \\ Ro_2{}^T S_{\omega 2} S_{\omega 2} \{\bar{d}^2_2(b)\} - Ro_1{}^T S_{\omega 1} S_{\omega 1} \{\bar{d}^2_1(b)\} \end{Bmatrix} \qquad \bar{a} = \begin{Bmatrix} a_{1x} \\ a_{1y} \\ a_{1z} \\ 0 \\ 0 \\ 0 \end{Bmatrix}$$

where, $\{\bar{d}^m_n(b)\} = \begin{Bmatrix} dis(x^m_n(b)) \\ dis(y^m_n(b)) \\ dis(z^m_n(b)) \end{Bmatrix}$: distance components in body frame.

APPENDIX B -MATRIX FOR HUMANOID DYNAMIC MODEL

$$C_T = \begin{bmatrix} -I & I & 0 & I & 0 & I & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & -I & I & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & -I & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & -I & I & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & -I & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & -I & I & 0 & 0 & I & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & -I & I & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & I & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & I \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I \end{bmatrix}$$

Where, I is 3×3 identity matrix which including components of x, y, and z

$B = C_T$

$S_R =$

$$\begin{bmatrix} -\bar{d}_{11} & \bar{d}_{12} & 0 & \bar{d}_{14} & 0 & \bar{d}_{16} & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & -\bar{d}_{22} & \bar{d}_{23} & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & -\bar{d}_{33} & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & -\bar{d}_{44} & \bar{d}_{45} & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & -\bar{d}_{55} & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & -\bar{d}_{66} & \bar{d}_{67} & 0 & 0 & \bar{d}_{610} & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{77} & \bar{d}_{78} & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{88} & \bar{d}_{89} & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{99} & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{1010} & \bar{d}_{1011} & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{1111} & \bar{d}_{1112} \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{1212} \end{bmatrix}$$

where $\bar{d}_{nn}$, is defined as, $\bar{d}_{nm} = \begin{bmatrix} 0 & -dis(z_n^m) & dis(y_n^m) \\ dis(z_n^m) & 0 & -dis(x_n^m) \\ -dis(y_n^m) & dis(x_n^m) & 0 \end{bmatrix}$.

(Components of distance vector in global frame)

n : index for mass center of each segment. m : index for each joint point.

$$C_A = \begin{bmatrix} I & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ -I & I & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & -I & I & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ -I & 0 & 0 & I & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & -I & I & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ -I & 0 & 0 & 0 & 0 & I & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & -I & I & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & -I & I & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & I & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & -I & 0 & 0 & 0 & I & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & I & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -I & I \end{bmatrix}$$

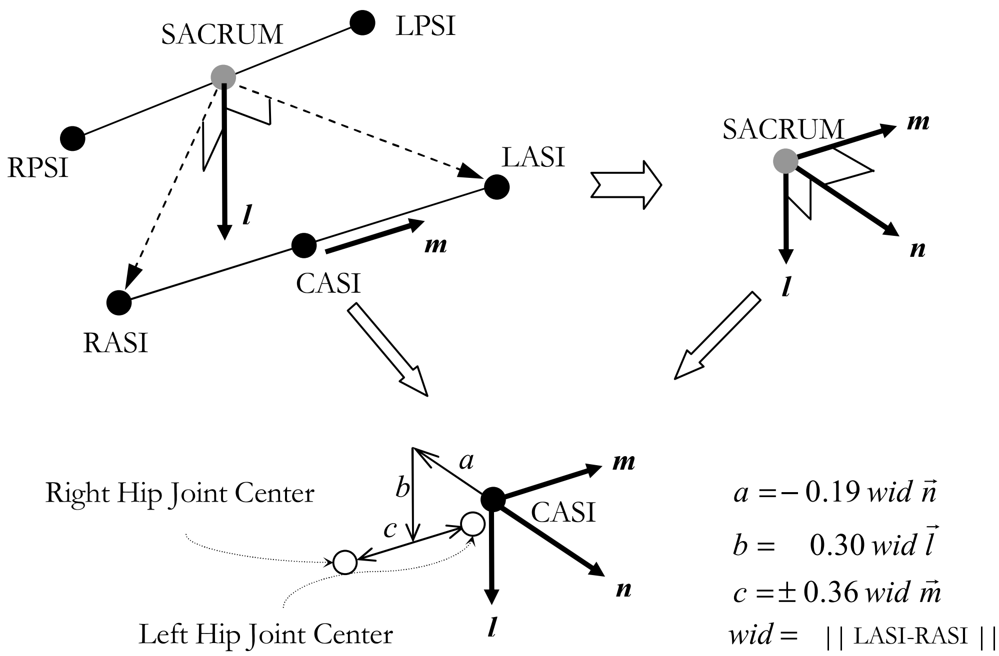
$$T_S =$$

$$\begin{bmatrix} \bar{d}_{11} & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ -\bar{d}_{12} & \bar{d}_{22} & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & -\bar{d}_{23} & \bar{d}_{33} & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ -\bar{d}_{14} & 0 & 0 & \bar{d}_{44} & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & -\bar{d}_{45}0 & \bar{d}_{55} & 0 & 0 & 0 & 0 & 0 & 0 & 0 \\ -\bar{d}_{16} & 0 & 0 & 0 & 0 & \bar{d}_{66} & 0 & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & -\bar{d}_{67}0 & \bar{d}_{77} & 0 & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{78} & \bar{d}_{88} & 0 & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{89} & \bar{d}_{99} & 0 & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & -\bar{d}_{610} & 0 & 0 & 0 & \bar{d}_{1010} & 0 & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{1011} & \bar{d}_{1111} & 0 \\ 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & 0 & -\bar{d}_{1112} & \bar{d}_{1212} \end{bmatrix}$$

$$\overline{S}_T = \begin{Bmatrix} Ro_1{}^T S_{\omega 1} S_{\omega 1} \left\{ \overline{d}_1^1(b) \right\} \\ -Ro_2{}^T S_{\omega 2} S_{\omega 2} \left\{ \overline{d}_2^2(b) \right\} + Ro_1{}^T S_{\omega 1} S_{\omega 1} \left\{ \overline{d}_1^2(b) \right\} \\ -Ro_3{}^T S_{\omega 3} S_{\omega 3} \left\{ \overline{d}_3^3(b) \right\} + Ro_2{}^T S_{\omega 2} S_{\omega 2} \left\{ \overline{d}_2^3(b) \right\} \\ -Ro_4{}^T S_{\omega 4} S_{\omega 4} \left\{ \overline{d}_4^4(b) \right\} + Ro_1{}^T S_{\omega 1} S_{\omega 1} \left\{ \overline{d}_1^4(b) \right\} \\ -Ro_5{}^T S_{\omega 5} S_{\omega 5} \left\{ \overline{d}_5^5(b) \right\} + Ro_4{}^T S_{\omega 4} S_{\omega 4} \left\{ \overline{d}_4^5(b) \right\} \\ -Ro_6{}^T S_{\omega 6} S_{\omega 6} \left\{ \overline{d}_6^6(b) \right\} + Ro_1{}^T S_{\omega 1} S_{\omega 1} \left\{ \overline{d}_1^6(b) \right\} \\ -Ro_7{}^T S_{\omega 7} S_{\omega 7} \left\{ \overline{d}_7^7(b) \right\} + Ro_6{}^T S_{\omega 6} S_{\omega 6} \left\{ \overline{d}_6^7(b) \right\} \\ -Ro_8{}^T S_{\omega 8} S_{\omega 8} \left\{ \overline{d}_8^8(b) \right\} + Ro_7{}^T S_{\omega 7} S_{\omega 7} \left\{ \overline{d}_7^8(b) \right\} \\ -Ro_9{}^T S_{\omega 9} S_{\omega 9} \left\{ \overline{d}_9^9(b) \right\} + Ro_8{}^T S_{\omega 8} S_{\omega 8} \left\{ \overline{d}_8^9(b) \right\} \\ -Ro_{10}{}^T S_{\omega 10} S_{\omega 10} \left\{ \overline{d}_{10}^{10}(b) \right\} + Ro_6{}^T S_{\omega 6} S_{\omega 6} \left\{ \overline{d}_6^{10}(b) \right\} \\ -Ro_{11}{}^T S_{\omega 11} S_{\omega 11} \left\{ \overline{d}_{11}^{11}(b) \right\} + Ro_{10}{}^T S_{\omega 10} S_{\omega 10} \left\{ \overline{d}_{10}^{11}(b) \right\} \\ -Ro_{12}{}^T S_{\omega 12} S_{\omega 12} \left\{ \overline{d}_{12}^{12}(b) \right\} + Ro_{11}{}^T S_{\omega 11} S_{\omega 11} \left\{ \overline{d}_{11}^{12}(b) \right\} \end{Bmatrix}$$

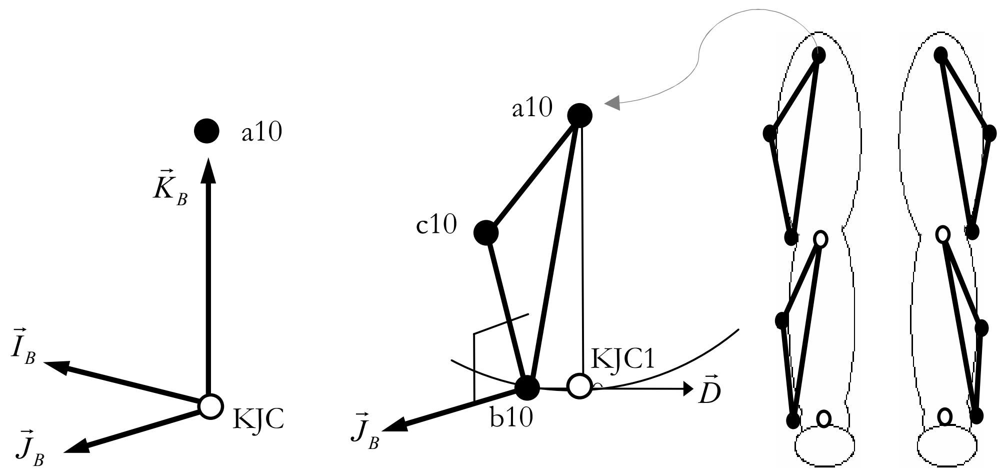
where, $\left\{ \overline{d}_n^m(b) \right\} = \begin{Bmatrix} dis(x_n^m(b)) \\ dis(y_n^m(b)) \\ dis(z_n^m(b)) \end{Bmatrix}$: distance components in body frame.

APPENDIX C –LOWER EXTREMITY JOINT CENTER POINT CALIBRATION.



2.8 Hip joint center calibration

(From the measured marker points of LASI, RASI, LPSI, and RPSI, the hip joint center (HJC), which is inside of body segment are calibrated based one the experimental data from previous study (Bell, 1990))



2.9 Knee and ankle joint center calibration

$\vec{J}_B$ is direction vector for forward direction which is obtained from the cross product of the vector from the point b10 to point a10 and the vector from the point b10 to point c10.

Vector D lies in the plane of the triangle including three points, (a10, b10, and c10) and is perpendicular to the direction of vector $\vec{J}_B$, and the vector from the point b10 to point a10. This vector D is not yet aligned with the vector connecting the point b10 and point KJC (but close).

By introducing temporal point KJC1, which is in the distance of half knee width from the point b10 in vector D direction, KJC2, which is much closer to KJC point can be obtained with the same procedure again.

Ankle Joint Center (AJC) can be calibrated with the same procedure used for KJC calibration.

DIFFERENCES IN THE POWER GENERATION OF LOWER EXTREMETIES BETWEEN HEALTHY EDLERLY FALLERS AND HEALTHY ELDERLY NON-FALLERS DURING GAIT

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ABSTRACT

Falls by older adults are the leading cause of injuries and trauma-related hospital admissions in the United States. Moreover, falls cause more than 90% of all hip fractures in the U.S. Twenty eight healthy elderly, over the age of 70 years, were recruited for normal gait test. Based on the self reported falling experiences, two groups were formed as fallers group (N=12), and non-fallers group (N=18). Spatiotemporal gait parameters and consumed mechanical works from lower extremities were calculated and compared between the groups to determine the unique gait characteristics of elderly fallers.

Narrower stride width and increased variability in generative work during knee extensor moment may be the greatest cause of falling for healthy elderly during walking. Wider stride walking with less variability in generative work during knee extensor moment can be good strategies for elderly to prevent possible future fallings.

INTRODUCTION

Human gait is an unstable mechanism which requires passive and active control to maintain stability. Many factors cause falls during gait combined with physiological condition of individuals. Falls are a major cause of decreased mobility and disability in the elderly.[1] Beyond mobility, fallings of elderly are the leading causes elderly death. The elderly, who represent 12 percent of the population, account for 75 percent of deaths from falls.[2] The number of elderly Americans death, which is associated with falls, is approximately 9,500 for each year.[3] By finding the unique gait characteristics, which elderly fallers have compared to elderly non-fallers, the method of preventing possible future elderly fallers can be designed.

Common kinematic gait pattern differences for the elderly fallers reported in previous studies are slow walk, reduced range of motion at ankle and hip, and extreme (too much or too little) stride width variance.[4-7] Chiba et al.[8] reported a significant decrease in toe clearance of fallers when compared with non-fallers. These reduction in speed and in ranges of motion could be related to the consumed mechanical works from each joint of body segments. McGibbon reported that joint muscle power and work were more closely associated with the physiology of human movement compared with spatial and temporal gait parameters.[9] Mechanical joint power is a sensitive parameter for human movement analysis because this represents kinetics and kinematics information simultaneously (moment and angular velocity). Rapid movements to recover from dynamic instability may be dependant upon the absolute power generating capacity of the muscles involved.[10]

The stability of gait based on the stride width differences was reported contradictory in the previous studies. In the study of Maki, increased stride width appeared to be predictive of falling while walking.[11] This result from Maki is contrary to the commonly assumed fact that wider stride increases stability.[12-14]

In this study, gait patterns of older adults (over 70 years of age) were compared between faller and non-faller groups in spatiotemporal parameters and in the consumed mechanical works from lower extremities. To date, nobody has examined the consumed mechanical works differences from hip and ankle in frontal plane to determine the gait characteristics of elderly fallers compared with non-fallers. The consumed mechanical works in the sagittal plane were calculated as in other previous studies, and the consumed works from the hip and the ankle in the frontal plane were calculated to monitor the lateral energy consuming pattern differences for healthy elderly fallers. Hip abduction strength for dominant leg side was measured to see the difference of hip muscle capability in lateral direction between the groups.

The obtained works were separated in several phases in the direction of works (generative or absorptive) and in the direction of moments (extensor moment, or flexor moment) based on the mechanical functions of each segments of lower extremities, and those divided works were compared between the groups. By these comparisons in more specific phases, consumed mechanical work comparisons represent the function of segments in locomotion.

This study explores the hypothesis that the healthy elderly fallers exhibit quantifiably different spatial characteristics and different consumed mechanical works than the healthy elderly non-fallers in average and in variability. To explore this hypothesis, the healthy elderly subjects were recruited to repeatedly ambulate across the test pass so that various gait parameters could be computed. For the calculations of consumed mechanical works, forward dynamics method combined with linearized feedback control system (FDMLFC), which had been developed from previous study, was used.[15] Findings of this study can be used as predicting parameters of future fallers, so prevention can be made before the falls happen.

METHODS

EXPERIMENTAL MEASUREMENTS

Twenty eight healthy seniors volunteered to participate in this study. The average age of subjects was 77 ± 4.6 years (mean $\pm$ S.D.), range 70−86 years; average height was 1.63 ± 0.10m, range 1.45−1.86m; average weight was 67.15 ± 11.90kg, range 47.2−90kg. Depending on the self reported experiences of falls, two groups were formed. Sixteen participants who had not experienced any falls within 12 months were classified as the non-fallers group. Twelve participants who had experienced falls within 12 months were classified as the fallers group. The protocol for the study was reviewed and approved by the Institutional Review Board (IRB) of Oregon State University.

Inclusion criteria were the following: who is 70 years of age or older, who can walk without assistance, and who has no physical or cognitive condition that would keep the individual from understanding or safely participating in the physical tests. Health conditions that could affect the gait patterns of participants were asked as exclusion criteria before the test (Appendix A-1). A Mini-Mental State Exam[16] was given to test each participant's memory and ability to understand and follow instructions.

Thirty-seven markers were attached to the participants, who were then asked to perform a series of trials in which they walked approximately 4 meters along a straight and level path. Participants were asked to walk naturally at their preferred, normal walking speed. A 3D motion capture system (Vicon Motion System, Oxford, United Kingdom) recorded the positions of all markers, and two force plates mounted on the middle of the path calculated the ground reaction forces synchronized with the motion capture data. The sampling frequency for the cameras was 60Hz, and 600Hz for the force plates. Participants were told to look straight ahead and not at the force plates on the ground. Lab technicians adjusted the participants' starting point to ensure that each foot made full contact with one force plate. This allowed each participant to get two complete foot contacts on two force plates without disrupting their natural gait pattern. Approximately 14 trials for each participant were conducted to achieve an average of 7 walking sets with two complete foot contacts.

Finally, the hip abduction strength was measured by using a dynamometer. Maximum isometric strength of the participant's hip abductor muscles was recorded from the hip of the dominant leg side. Dominant leg was decided by asking the participant to walk forward several times. In case the starting leg had consistency, that starting leg was set as a dominant leg. And in the case of no consistency in the starting leg, the participant was asked to kick the small object and the leg used for kicking was set as a dominant leg. Hand-held dynamometer was placed against the outside of one of the participant's thighs, two inches above the knee. The participant was asked to use his or her hip muscles to push his or her thigh sideways against the dynamometer. In the same time, a research staff member manually applied an opposing force to the dynamometer so that the participant's thigh would not move. Three pre-trials were conducted for warming up purposes, and three real trails followed. The maximum force between three trails was taken as hip abduction force.

DATA ANALYSIS

The spatiotemporal gait parameters were obtained from kinematic information and force plate data. Mean longitudinal distance between the ankle joint centers of the same foot during initial ground contact of sequential stances was recorded for stride length. Stride width was measured from mean lateral distance between ankle joint centers of both feet during initial sequential ground foot contacts. Gait speed was obtained from the measured stride length and spent time for one stride. The obtained spatiotemporal gait parameters were divided by the leg length of participants to eliminate the factor coming from different leg length.

Instead of traditional inverse dynamics method, forward dynamics method combined with linearized feedback control system (FDMLFC) was used in this study for joint moment calculation. In FDMLFC method, a 3D humanoid dynamic model was developed with twelve rigid bodies. The required inputs for humanoid dynamic model were obtained from non-linear feedback control system. The reference angles, (desired angles) which humanoid dynamic model was forced to track were obtained from 3D motion capture system. A detailed method and procedure was reported in the previous study of Ko et al.[15]

The FDMLFC method calculated the moments with respect to global frame. Those moments in global frame were translated to each sequential body frame, which the rotation of each segments was based on. The obtained moments as control efforts were used for mechanical joint power calculations by multiplying them with joint angular velocity.

Consumed mechanical works during one stance were obtained from integration of obtained mechanical powers. Works from joints of hip, knee, and ankle were stored and separated based on the phase of moments (extensor and flexor) and the phase of works (absorptive and generative). Figure 3.1 and Figure 3.2 show the kinematic and kinetic plots for a representative subject in sagittal plane for hip, knee, and ankle, and in frontal plane for hip and ankle. The consumed mechanical works in sagittal plane from lower extremities consisted of the following: generative work during hip extensor moment (H1) in beginning of stance, absorbing work during hip flexor moment (H2) in middle of stance, and generative work during hip flexor moment (H3) in terminal stance; first absorbing work during knee extensor moment (K1), generative work during knee extensor moment (K2), generative work during knee flexor moment (K3), and second absorbing work during knee extensor moment (K4); absorbing work during ankle dorsal flexor moment (A1), absorbing work during ankle plantar flexor moment (A2), and generative work during ankle plantar flexor moment (A3). The consumed mechanical works in frontal plane from hip and ankle consisted of the following: absorbing work during hip abductor moment in first half of stance (HF1), and generative work during hip abductor moment (HF2); absorbing work during ankle inverter moment (AF1), absorbing work during ankle evertor moment (AF2), and generative work during ankle evertor moment (AF3). The obtained mechanical works were normalized by the participant's mass and stride length so that the obtained parameters could be compared between the groups.

The consumed mechanical works which had been normalized were compared between the two groups of the non-fallers and fallers as average and variability (mean standard deviation) of both groups. Depending on the stance leg, the consumed works also compared between the work from dominant leg stance and the work from non-dominant leg stance. A two way ANOVA test was performed to detect the group differences of the consumed work with two factors including fall (fallers and non-fallers) and limb (from dominant leg stance and from non-dominant leg stance). *P*-value of less than 0.05 was selected to indicate significant differences.

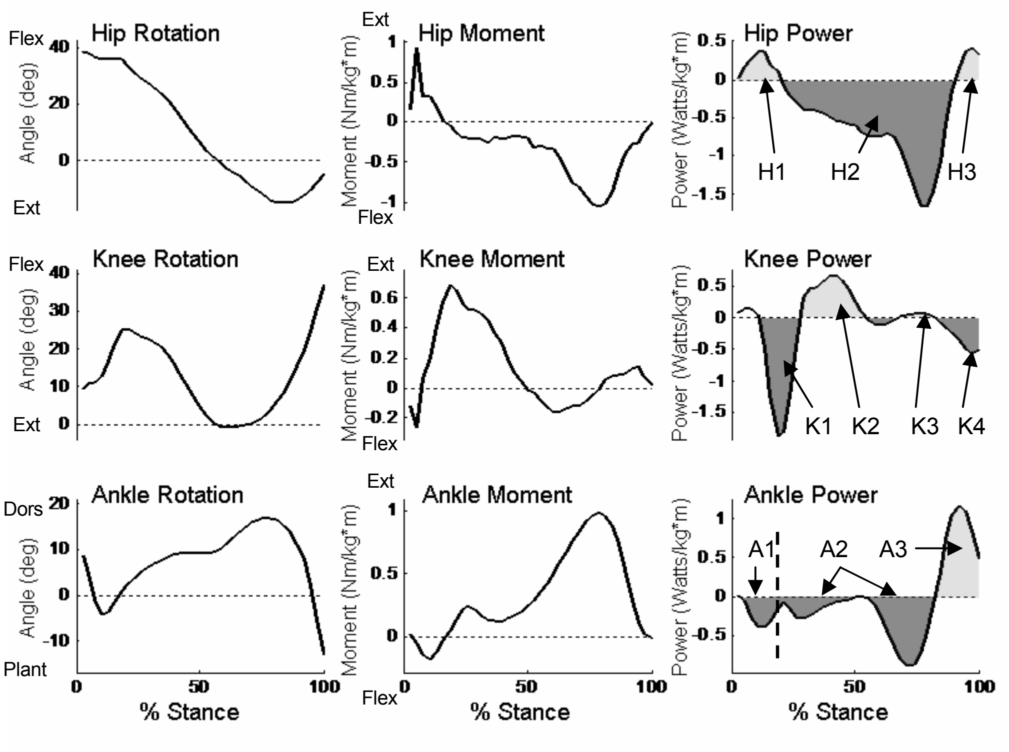

Figure 3.1 Angle, moment, and mechanical powers from lower extremities in sagittal plane

H1 : Hip extensor generation
H3 : Hip flexor generation
K1 : Knee extensor absorption (during 0~50% of stance)
K2 : Knee extensor generation
K3 : Knee flexor generation
K4 : Knee extensor absorption (during 50~100% of stance)
A1 : Ankle dorsal flexor absorption
A2 : Ankle plantar flexor absorption
A3 : Ankle plantar flexor generation

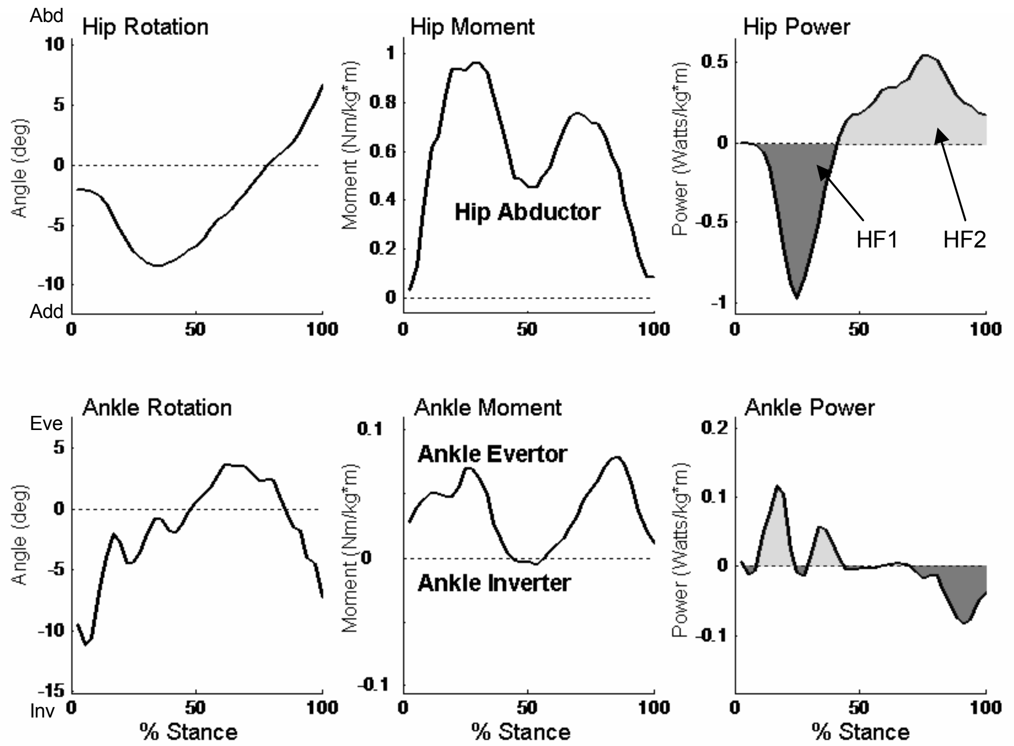

Figure 3.2 Angle, moment, and mechanical powers from the lower extremities of hip and ankle in frontal plane

HF1 : Hip abduction absorption (during 0~50% of stance)

HF2 : Hip abduction generation

RESULTS

As shown in Table 3.1, stride width significantly differed among groups of fallers and non-fallers. Gait speed, stride length and measured leg abduction force did not show significant differences between the groups.

The average and variability of consumed work were calculated in four different groups based on falling experience (fallers and non-fallers) and the leg used for stance (dominant leg and non-dominant leg). No significant interaction between factors (falling and dominant leg) was found except the variable H2 (absorbing work during hip flexor moment) in average. The effects of falling factor were not significant to the groups of dominant leg and non-dominant leg in average variable of H2. No significant differences were detected for fallers compared with non-fallers in sagittal and frontal plane as shown in Table 3.2 and Table 3.3. The variability of consumed work in individual trials of each subject was computed to see the consistency of the consumed works of each subject from lower extremities. Comparisons of variability between the groups were conducted and shown in Table 3.4 and Table 3.5 for sagittal plane and frontal plane respectively. Significantly higher variability in K2 (generative work during knee extensor moment) was found for fallers compared with non-fallers (p = 0.04).

Table 3.1 Average comparisons of spatiotemporal data and leg abduction forces between the subject groups of fallers and non-fallers

	Non-fallers (Mean ± S.D.)	Fallers (Mean ± S.D.)	*p* value
Gait speed (m/s)[a]	1.267 ± 0.212	1.366 ± 0.259	0.42
Stride length (m)[b]	1.576 ± 0.256	1.597 ± 0.262	0.94
Stride width (m)[c]	0.138 ± 0.036	0.109 ± 0.025	*0.025*
Hip Abduction (kg)	27.631 ± 7.564	25.958 ± 5.121	0.515

[a, b, c] Parameters are normalized with participant's leg length.

Table 3.2 Average comparisons of consumed works from lower extremities in sagittal plane between the groups of non-fallers and fallers

Mechanical work (J / m kg)	Non-fallers (Mean ± S.D.)	Fallers (Mean ± S.D.)	*p* value
H1	0.0169 ± 0.0075	0.0147 ± 0.0092	0.48
H3	0.0016 ± 0.0009	0.0016 ± 0.0012	0.91
K1	-0.0121 ± 0.0067	-0.0171 ± 0.0098	0.1
K2	0.0066 ± 0.0042	0.0096 ± 0.0059	0.13
K3	0.0072 ± 0.0050	0.0106 ± 0.0073	0.19
K4	-0.0042 ± 0.0048	-0.0027 ± 0.0034	0.29
A1	-0.0046 ± 0.0044	-0.0047 ± 0.0030	0.94
A2	-0.0387 ± 0.0143	-0.0309 ± 0.0118	0.12
A3	0.0301 ± 0.0107	0.0271 ± 0.0069	0.39

H1 : Hip extensor generation
H3 : Hip flexor generation
K1 : Knee extensor absorption (during 0~50% of stance)
K2 : Knee extensor generation
K3 : Knee flexor generation
K4 : Knee extensor absorption (during 50~100% of stance)
A1 : Ankle dorsal flexor absorption
A2 : Ankle plantar flexor absorption
A3 : Ankle plantar flexor generation

Table 3.3 Average comparisons of consumed works from hip and ankle in frontal plane between the groups of non-fallers and fallers

Mechanical work (J / m kg)	Non-fallers (Mean ± S.D.)	Fallers (Mean ± S.D.)	*p* value
HF1	-0.0165 ± 0.0082	-0.0177 ± 0.0070	0.69
HF2	0.0129 ± 0.0058	0.0145 ± 0.0051	0.47
AF1	-0.0013 ± 0.0013	-0.0011 ± 0.0016	0.5
AF2	-0.0033 ± 0.0036	-0.0036 ± 0.0047	0.96
AF3	0.0032 ± 0.0027	0.0059 ± 0.0058	0.16

All Parameters are normalized with participant’s weight and stride length.

HF1 : Hip abduction absorption (during 0~50% of stance)
HF2 : Hip abduction generation
AF1 : Ankle inverter absorption
AF2 : Ankle evertor absorption
AF3 : Ankle evertor generation

Table 3.4 Variability comparisons of consumed works from lower extremities in sagittal plane between the groups of non-fallers and fallers

Mechanical work (J / m kg)	Non-fallers (Mean ± S.D.)	Fallers (Mean ± S.D.)	*p* value
H1	0.0073 ± 0.0019	0.0058 ± 0.0028	0.13
H3	0.0007 ± 0.0003	0.0007 ± 0.0005	0.85
K1	0.0041 ± 0.0026	0.0045 ± 0.0019	0.68
K2	0.0024 ± 0.0011	0.0034 ± 0.0010	*0.04*
K3	0.0032 ± 0.0017	0.0028 ± 0.0016	0.59
K4	0.0027 ± 0.0017	0.0019 ± 0.0016	0.24
A1	0.0018 ± 0.0015	0.0019 ± 0.0015	0.82
A2	0.0070 ± 0.0027	0.0061 ± 0.0018	0.34
A3	0.0065 ± 0.0023	0.0063 ± 0.0025	0.85

H1 : Hip extensor generation
H3 : Hip flexor generation
K1 : Knee extensor absorption (during 0~50% of stance)
K2 : Knee extensor generation
K3 : Knee flexor generation
K4 : Knee extensor absorption (during 50~100% of stance)
A1 : Ankle dorsal flexor absorption
A2 : Ankle plantar flexor absorption
A3 : Ankle plantar flexor generation

Table 3.5 Variability comparisons of consumed works from hip and ankle in frontal plane between the groups of non-fallers and fallers

Mechanical work (J / m kg)	Non-fallers (Mean ± S.D.)	Fallers (Mean ± S.D.)	*p* value
HF1	0.0029 ± 0.0014	0.0032 ± 0.0011	0.57
HF2	0.0027 ± 0.0011	0.0027 ± 0.0012	0.93
AF1	0.0007 ± 0.0005	0.0007 ± 0.0010	0.81
AF2	0.0007 ± 0.0005	0.0006 ± 0.0005	0.54
AF3	0.0016 ± 0.0022	0.0013 ± 0.0006	0.54

HF1 : Hip abduction absorption (during 0~50% of stance)
HF2 : Hip abduction generation
AF1 : Ankle inverter absorption
AF2 : Ankle evertor absorption
AF3 : Ankle evertor generation

DISCUSSION

Finding out the unique gait patterns of healthy elderly fallers compared to healthy elderly non-fallers is the main goal of this study. Gait analysis based on comparisons of gait parameters between the non-fallers group and the fallers group was performed. Two healthy elderly groups were formed based on the falling experiences the participants had reported. No physical and mental problems which might affect the normal locomotion had been reported in either group. The consumed mechanical works from lower extremities during normal walking tests were compared between the groups. Hip abduction force and spatiotemporal data including stride length, stride width, and speed of gait were also computed and compared between the groups.

The elderly fallers group in this study walked with narrower stride width compared to the elderly non-fallers group. This result matches well with the commonly assumed fact that wider stride increases stability.[12-14] Narrower stride width could cause instability in both single support and double support phases. In single support phase, because the COM (center of mass) is closest to the edges of the base of support, small movement in upper body from inside to outside in lateral direction can make that period the most dangerous. In the double support phase, the narrower stride width also decreases the lateral direction stability for the same reason. Maki reported that increased stride width appeared to be predictive of falling while walking.[11] This result of Maki's study is contrary to the result of this study. Elderly fallers in his study could have chosen the wider stride width for fear of falling. But again this wider stride walking could have caused other fallings because of wider movement of COM. The previous study reported that elderly fallers have difficulty in controlling lateral stability.[17, 18] The elderly fallers in this study might have chosen narrower stride width for straight gait with less control in lateral direction.

Stride length, gait speed, and hip abduction strength between both groups did not show significant differences. Many previous studies reported that elderly fallers walk slower compared to elderly non-fallers.[4-7] Similarly, in the study of Kerrigan, comfortable gait speed of elderly fallers was slower compared with comfortable gait speed of elderly non-fallers.[19] These inconsistencies of stride length and gait speed between studies seems to be due to the different criteria for forming the groups of fallers and non-fallers. In the study of Gunter, no difference was found in hip abduction strength between the fallers group and non-fallers group, and that result was the same with this study.[20]

Prior studies reported reductions in plantar flexion and plantar flexor powers in healthy elders compared with young adults.[21-23] Talbot et al. reported that falls increased with age.[24] Considering there are more falls from healthy elderly compared with healthy young adult, this reduction of ankle power in elders could be exaggerated in the elderly fallers compared to the elderly non-fallers. But in this study, no significant difference in ankle work was found in neither generative and absorbing way during plantar flexor moment between the groups as shown in Table 3.2.

In the study of Kerrigan, the certain kinetic reductions that persisted at both comfortable and fast walking speeds in elderly fallers compared to elderly non-fallers were found.[19] These results can be interpreted that the decrease in gait speed may not be the reason for kinetic reductions in elderly fallers compared to elderly non-fallers.

The consumed mechanical works from lower extremities did not show the significant differences between the groups in a sagittal plane and frontal plane. Same stride length, same abilities for the hip abduction force generation, and no functional limitations of both groups might cause the same kinetic patterns of mechanical work consumptions in sagittal plane. The difference in stride width though did not change the consumed works of the hip and ankle in frontal plane in this study. This result indicates that hip power from abductor moment and ankle power from ankle evertor and inverter moment during stance are not much related with stride width. Those powers of the hip and ankle in frontal plane could be more associated with the distance between hip joint center and COM (Center of mass) of upper body.

Although the study of Brach[7] showed that the extreme step width variability (small or large variances) is associated with the falls of elderly, the step width variability in this study did not show significant difference between the groups, but the variability of generative work during knee extensor moment (K2) was significantly higher for the fallers group compared with non-fallers group ($p = 0.04$). This finding of variability could indicate the consistency of gait. The increased variability may indicate an increased risk of falling during walking as a result of errors in control of foot placement and/or center-of mass displacement.[11]

In this study, the consumed mechanical work comparisons were performed between the groups only for the lower extremities. Pelvic or upper body kinematics and kinetics were not included for the analysis, though those upper body parameters might have been useful for describing the gait patterns of the elderly fallers. A gait test with a larger sample size would have provided more reliable results revealing the gait pattern characteristics of the elderly fallers. In the future study, by forming the fallers group based on the falling test instead of the reported information of participant's falling experiences, the consumed mechanical work differences for the elderly fallers can be better understood. The consumed mechanical works applied to execute the response movement of the real falling test could provide a better idea for determining future elderly fallers.

The consumed mechanical works used in this study for kinetic gait parameters represent the muscle activities compared to range of angles, peak moments, and peak powers of traditional gait studies. The mechanical works consumed from hip and ankle in frontal plane during stance phase were not shown from previous studies, though the functions of hip and ankle in frontal plane during stance are important to maintain the lateral balance. Although no frontal work differences were found in this study, the mechanical consumed works in frontal plane could be used to identify the gait patterns of fallers or sideway-fallers in future studies.

Increased variability in generative work during knee extensor moment (K2) and reduced stride width are significant differences a healthy elderly fallers have compared with healthy elderly non-fallers. Walking with wider stride width may be a better strategy for healthy elderly to prevent possible future fallings.

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APPENDIX

APPENDIX A – HEALTH CONDITION AND FALLING HISTORY QUESTIONNAIRE

Health Condition

Have you ever had:	**Yes**	**No**
Balance problems		
Frequent dizziness		
Stroke or other brain injury		
Epilepsy or seizures		
Heart problems or heart attack		
Hip or knee replacement		
Do you currently have:	**Yes**	**No**
Osteoporosis		
Muscle injury or tendonitis		
Diabetes		
Arthritis or severe joint pain		
Parkinson's disease		
Respiratory disease (e.g. asthma)		
Uncorrected vision problems		
Any other medical conditions that affect your ability to walk or keep your balance? If Yes, please describe:		
Do you take sedatives or antidepressants?		
Do you use an assistive device to walk?		

Falling History

A "fall" is when you unintentionally come to rest on the ground or some other lower level, except if you faint or are hit by a violent blow

1. **Have you experienced a fall during the past 12 months?**
   (Circle one answer, then follow the arrow to next question)

   No

   YES

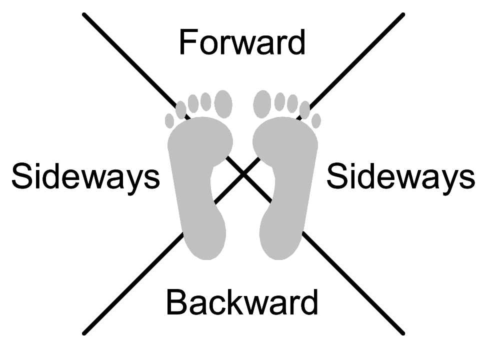
2. **How many times did you fall in the past 12 months?**
   (Mark one box)

Once	
More than once	

We consider the direction of a fall to be the direction that your body moved relative to your feet as you fell *(see picture below)*.

3. **In what direction (or directions) have you fallen in the past 12 months?**
   (Mark each box that applies)

Sideways	
Forward	
Backward	



GENERAL CONCLUSION

Falls of elderly are a major source of elderly injuries, and even deaths. Finding out the unique gait patterns healthy elderly have depending on their different falling experiences is the main goal of this study.

The identifiers for side way fallers were determined in the purpose of identifying the fallers who could have more chances for hip fractures. The new forward dynamics method was introduced as an alternative of inverse dynamics method in calculations of kinetic information for human gait movement. The consumed mechanical works from lower extremities were compared in sagittal plane and frontal plane to find the different energy consumption for healthy elderly fallers compared to health elderly non-fallers.

The findings from the first study indicate that stride width is associated with side-fall incidence among older fallers walking at a "normal," self-selected walking speed. Specifically, side-fallers, who walked with narrower strides compared to those who fell in other directions (forward, backward, straight down), may not be adapting their gait to compensate for deficits in medio-lateral stability.

The new method of three dimensional kinetic analyses was developed in the second part of this study. Forward dynamics method combined with linearized feedback control system (FDMLFC) was introduced. In calculation of the required kinetics from the known human movements, FDMLFC can be used as an alternative method of inverse dynamics without error propagation and without using double numerical differentiation. The result from numerical simulation with FDMLFC was compared with the result from inverse dynamics for verification purposes. The similar pattern in moments illustrates that the developed humanoid dynamic model properly represents the human motion dynamics. A satisfactory tracking simulation demonstrates that the non linear feedback control system was properly designed.

In the third part of this study, comparisons in gait parameters such as spatiotemporal gait parameters, and the consumed mechanical work from lower extremities in a sagittal and frontal planes were performed between the healthy elderly groups of fallers and non-fallers to find the gait pattern of healthy elderly fallers. The consumed works from lower extremities did not show the significant differences between the groups in a sagittal plane and frontal plane. Same stride length, same abilities for hip abduction strength generation, and no functional limitations of both groups might result in same kinetic pattern of mechanical work consumption. Increased variability in generative work during knee extensor moment (K2), and reduced stride width are significant differences healthy elderly fallers have compared with healthy elderly non-fallers.

Stride width, which is a relatively simple gait parameter, was determined as a variable showing differences between the groups of two separated studies. Elderly fallers showed narrower stride width compared to elderly non-fallers. Elderly side way fallers showed narrower stride width compared to elderly other direction fallers. Based on these two results of two studies, walking with a stride width which is not narrow could be an ideal choice for elderly to prevent future falls and hip fractures.

The analysis of whole body movement can be performed in a future study. Pelvic or upper body kinematics and kinetics were not included for the analysis, though those upper body parameters might have been useful for describing the gait patterns of the elderly fallers. A gait test with a larger sample size would have provided more reliable results revealing the gait pattern characteristics of the elderly fallers. In the future study, by forming the fallers group based on the falling test instead of the reported information of participants' falling experiences, the consumed mechanical work differences or spatiotemporal gait differences for the fallers can be better understood. The analysis about the response movement of real falling test could provide a better idea about who has a greater chance to be a faller in the future.