Master's Thesis
석사 학위논문

Numerical analysis of ZMP based biped walking model with different stiffness of ankle and toe joint

Dong Woo Nam (남 동 우 南 東 佑)

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Co-advisor : Professor Jong Soo Lee

by

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Department of Robotics
DGIST

A thesis submitted to the faculty of DGIST in partial fulfillment of the requirements for the degree of Master of Science in the Department of robotics engineering. The study was conducted in accordance with Code of Research Ethics[1]

1(month). 8(day). 2016(year)

Approved by

Professor Sang Jun Moon (Signature)
(Advisor)

Professor Jong Soo Lee (Signature)
(Co-Advisor)

[1] Declaration of Ethical Conduct in Research: I, as a graduate student of DGIST, hereby declare that I have not committed any acts that may damage the credibility of my research. These include, but are not limited to: falsification, thesis written by someone else, distortion of research findings or plagiarism. I affirm that my thesis contains honest conclusions based on my own careful research under the guidance of my thesis advisor.

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Accepted in partial fulfillment of the requirements for the degree of Master of Science

1(month). 8(day). 2016(year)

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ABSTRACT

Natural walking has large needs in many ways, for example, amputees that lost their lower limb. Amputees cannot walk well with their conventional prosthesis because of lack of muscle. This causes not suitable motion of ankle and toe joint during their walking. To walk naturally like normal people, actuation of muscle is needed because this generates variable joint stiffness of ankle and toe joints. Joint stiffness, which is the most important component that can control the position of foot and toe during walking. Repeatedly, with joint stiffness change, the movement of human body during walking can be changed in form of natural walking like normal people. In this study, the change of whole body movement during walking through walking direction will be dealt with ZMP based biped walking model. In the condition of fixed joint angle change, joint torque change can be dealt with only the change of joint stiffness. Also in terms of acceleration of human body can be changed when varied torque is considered. Finally, by variation of acceleration term, the movement of human body is changed similar to real human motion.

Keywords: ZMP based biped walking model, joint stiffness, joint torque

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Ⅰ. INTRODUCTION

1.1 Needs and understand of natural walking

Natural walking of human has become important issue in many reasons. For example, walking of lower limb amputees with conventional prosthesis that cannot mimic the walking of normal human is largely considered problem [1]. One of the problem is, amputees who lost his or her leg by gangrene and many other accidentsAm [2] consumes more than at least 10% to at most 30% of metabolic energy when they are walking [3]. Also, unnatural walking of amputees with conventional passive prosthesis was revealed by active prosthesis ankle research [4] by measuring the joint angle change during walking that seems not same form; when comparing with the joint angle change of sound limb, it shows smaller angle change that less than 2 degree at the ankle joint, less than 10 degree at the toe joint. All of these problems are caused by the lack of function of calf muscle and whole function of ankle joint [5]. By these kind of reasons, natural walking of human should be researched largely. Detailed figure is shown in Figure 1-a and b. Figure 1-a shows conventional passive type prosthetic ankle and schematic of amputee that equipped prosthesis. Figure 1-b shows penalty of joint angle change of ankle and toe joint when prosthesis is equipped as mentioned before.

Procedure of normal human walking should be understood for realizing of natural walking. Also this can help understanding of movement of ankle and toe joint. Walking phase of normal human can be divided in two phase; Detailed picture of walking cycle is shown in Figure 1-c. Stance phase (62% of whole cycle) and swing phase (38% of whole cycle) [6]. At the beginning of stance phase, heel touches the ground and foot flat, which means the perfectly landing of foot is occurred. After foot flat, loading and dorsiflexion of ankle joint is beginning. This phase is called mid-stance phase. In the terminal stance phase, toe adduction and plantarflexion of ankle are beginning and ready to kick the ground off. Finally, acceleration of foot is occurring in pre-swing phase, which is the last phase of stance phase. In this phase, thrust for kicking off the ground is generated and stance phase is over. After toe off, standing phase is changed to swing phase. During swing phase, foot comes back to its initial position and ready to hit the ground again. For connecting the ground with plantar again, ankle should be dorsiflexed. When heel strikes the ground, swing phase is changed to stance phase again and new walking cycle is beginning.

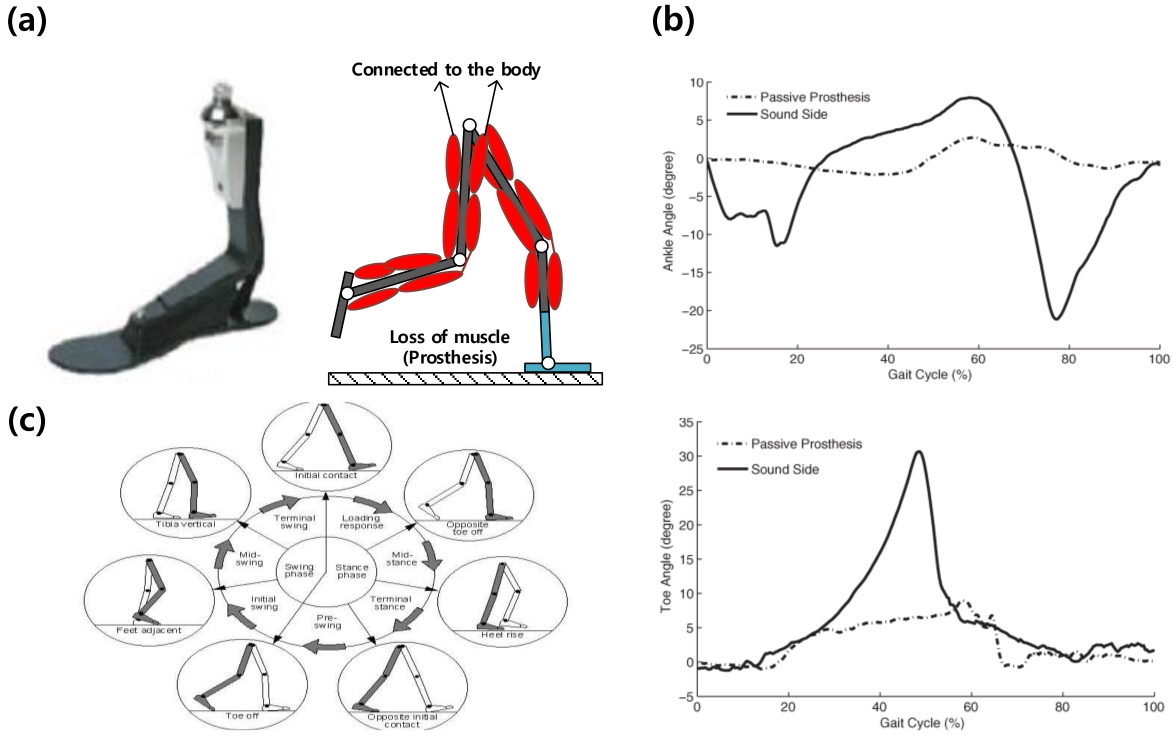

Figure 1 (a) Natural walking, which means similar as possible as human walking is needed for variable areas. For instance, conventional prosthesis design for amputees (left) [1]. Right figure shows the schematic of real human model with conventional prosthesis. However, still not be benefited by natural walking, like conventional prosthesis. Because of its simple structure and lack of mechanical assistant such as joint stiffness that largely affected by muscle, amputees with this kind of prosthesis face to trouble when they are walking. For instance, (b) shows the different form of ankle angle during whole walking cycle of amputee who uses the conventional passive type prosthesis [4]. In conventional prosthesis, both of ankle and toe joint has no function during walking, as (b) shows; both of ankle and toe joint of prosthesis side has little movement in contrast to sound sides of lower limb. Considering of natural walking of human is largely needed for this kind of lower limb amputees that cannot walk well as normal people. Before considering about natural walking, walking cycle of human should be understood, especially detailed motion during walking cycle. (c) The walking cycle of human, which is essential for understand of natural walking. This cycle can be divided in two steps; stance phase and swing phase. Stance phase starts with heel strike that heel touches the ground. After heel strike, whole foot contacts with the ground. When body moves enough, heel loses its contact with the ground and other side heel touches the ground. When ankle shows maximum plantar flexion, toe kicks off the ground to lift off and this is called as toe off.

1.2 Role of ankle and toe joint during walking

Foot charges important role during walking [6, 7] and ankle joint is the hinge type joint that links shank and foot [6]. Especially, addition of foot has advantage of reducing joint torques than the case of walking without foot [7]. Also ankle joint helps the movement of foot during walking by its two types of movement; dorsiflexion, which means the movement of ankle joint that pull the foot up to the shank, the other is plantarflexion which means the movement of ankle joint that push the foot down to the ground. By this movement, net positive work is generated at the ankle joint and this causes increase of walking velocity [3]. Moreover, the largest burst of work through all lower limb joint is produced at the ankle joint during walking [8]. Therefore, addition of ankle joint should not be omitted when designing the biped walking, especially when considering about natural walking of human.

When foot kicks the ground off at the end of stance phase, exist of toe joints has also advantages than the case of rigid foot without toe joint, for instance, walking speed can be increased [9-11] or increase of stability during walking because of extension of double support phase duration [12]. Some studies shows these phenomenon, especially humanoid studies such as [9]. In [9], addition of toe joint on the humanoid robot shows the reduce movement range of knee joint, which means reduce of joint torque and energy during walking, as mentioned before. Therefore, consideration of toe joint is also essential for natural walking. In next chapter, joint stiffness, which is the most important thing to generate the joint torque and joint position during walking.

1.3 Importance of joint stiffness

Both of ankle and toe joint helps human like natural walking. Also these joints should be their own position during walking. To make this joints keep their position during walking cycle, joint stiffness should be dealt. Joint stiffness, which commonly dealt in biomechanics [13] has big interest because by this variable, adjustment of joint positions or joint torque is available [14]. This value can be classified as quasi-stiffness again [13, 15] that means an ability of the system to endure externally imposed displacements [13]. The value of these kind of quasi-stiffness is able to be used as an index for selection and adjustment of the spring coefficient of prosthesis [15]. Especially, these kind of adjustment is important because the end effector; in this study, means foot or toe segments should be maintained as desired position [14]. So, to maintain foot and toe keep their position during walking, ankle and toe join stiffness adjustment is essential. In next chapter, the stiffness of these two joint will be described.

1.4 Previous study and hypothesis

Human walking procedure that mentioned in the end of the first chapter can be satisfied by the variable stiffness of lower limb joint during walking which is caused by actuation of muscle [16]. Based on this theory, Huang *et al* presented the influence of ankle joint stiffness during walking [17] and the relationship of walking speed and joint stiffness of ankle and toe joint [10] with passive walking model that consist with only toe, foot and leg segments on the tilted ground. Also in [10], advantages from compliance foot segment were dealt. Recently, based on [10], Jinying *et al* [4, 18] developed the active type prosthetic ankle with 2 SEAs (series elastic actuator) which is shown in Figure 2. In [18], only designing of active type prosthesis was dealt and clinical test on amputee was done in [4]. No studies about toe stiffness were comprehensively done, even in [10], so Jinying measured toe joint angle and moment changes during walking in constant speed and adopt to the SEA. With this prosthesis prototype, amputee could walk like normal human. Moreover, they found suitable ankle and toe joint stiffness value through their prototype model during average speed walking of amputee. In figure 2, schematic of active type prosthesis and the results of walking. The shape of ground reaction force during whole walking cycle was main observation of their study. However, in the research of Jinying [4, 18], only prototyping was exist, numerical modeling or proof of walking model was not dealt. Without numerical modeling, adaptation to variable body size of different people can be hard. Moreover, Jinying did their experiment in fixed condition; fixed walking speed, only one amputee. Especially, with only one amputee, fixed body dimension can be interacted as limitation; as mentioned before, variation of body dimension cannot be considered. Although Huang [10] studied about ankle and toe joint stiffness, it was done with simplified passive walking model; walking model with only one leg, foot and toe segment, so it was not too similar as walking of real human. By this reason, study about human like walking model for variable joint stiffness value during walking is reasonable.

For representation of natural walking analysis, in this study, built hypothesis like below. As joint torque can be represented in both of terms of acceleration of body mass (and inertia of each lower limb segments in detail) and joint stiffness. If joint angle during walking does not change, joint torque can be changed by change of joint stiffness. Then, by changed joint torque, acceleration of body will be changed (inertia of lower limb segment will not be changed because this value is unique value of each lower limb segments). This means the movement of body mass can be changed. Biped walking model is selected as ZMP based walking model endnote that has a simplified trajectory of body mass during walking. By the change of joint stiffness, observation of changed walking trajectory of body mass is the purpose of this study. By this study, more natural walking condition can be found through both of numerical analysis and walking simulation.

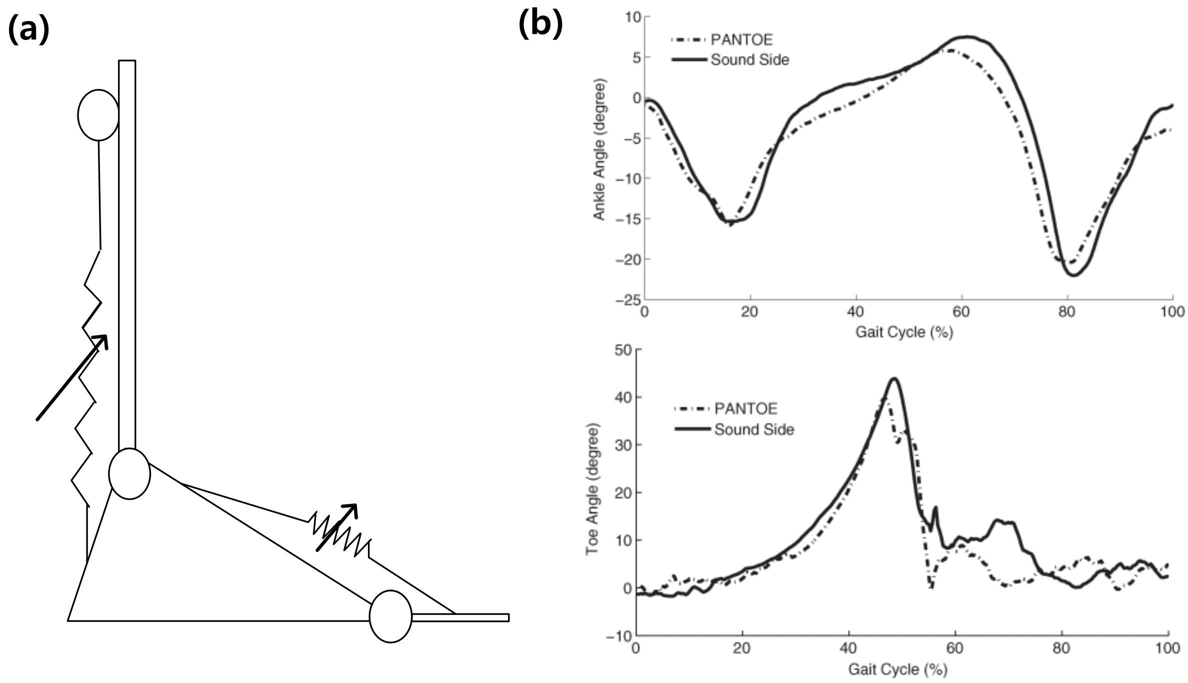

Figure 2 During walking, stiffness of each joint of lower limb is changed regularly by function of muscle as mentioned before, which was revealed by Frigo [16]. Also Huang [10] revealed the relationship of walking speed and ankle stiffness during walking. Based on this theory, Jinying *et al* developed the stiffness adaptable ankle prosthesis which called PANTOE with two SEA [4, 18] and (a) shows the schematic of PANTOE. This makes amputee walks more naturally like normal people and the results of prosthesis walking test is shown in b). In (b), unlike Fig. 1-(b), with PANTOE, which was developed in this study [4, 18], angle change during walking cycle both of ankle and toe joint of prosthetic limb shows almost same forms with sound lower limb. With joint stiffness variable prosthesis, more natural walking can be observed. This always means the importance of variable joint stiffness during walking for natural walking of amputees by keeping own position of each joint. In this study, suitable joint stiffness of ankle and toe joint for stable walking will be dealt in fixed walking speed by walking model simulation based on ZMP biped walking model. By this study, suitable joint stiffness that changes variety can be found in given walking speed in variable body dimension condition of walking model.

Ⅱ. Material and methods

2.1 Kinematic description of walking model

Dimension of model was obtained by following procedures. Basically, model was designed in biped walking model. Shank and thigh segments was decided by proportion of body segments by Winter [19]. Based on this information, dimension of walking model decided with assumption that this model was calculated from a 1.7m of height and 63kg of weight human, which was used as average data in [20]. With this calculation, foot length and shank length are respectively 0.2584m (25.84cm, 0.152 times of height) and 0.417m (41.7cm). In case of height of ankle is 0.068m (6.8cm). Length of thigh is same with shank, 0.417m. In case of foot, however, should be improved from flat foot model without toe to arched foot with toe. Scale of toe was determined from the data of the foot size of normal people from SizeKorea [21]. This data is formed with approximately 3000 healthy Korean men and women (1500 men and 1500 women) whose age is between 20 and 59. More detailed information is explained in table 1. In this data, averaged representative scale of human toe is included. With these detailed dimension of toe, dimension of toe could be calculated. The result of calculation is 0.068m, which is approximately 0.27 times of foot length.

Part name	***Link name***	***Ratio (vs height)***	***Length (cm)***
Foot	l_{foot}	0.152	25.84
Toe	l_{toe}	0.04	6.8
Bare foot	l_{bf}	0.051	8.67
Fore foot	l_{ff}	0.101	17.17
Foot without toe	l_{ft}	0.112	19.04
Ankle height	l_{ankle}	0.04	6.8
Shank	l_{shank}	0.245	41.7
Thigh	l_{thigh}	0.245	41.7

Table 1 Detailed body link length of walking model that followed the ratio of Winter [19]. In case of toe length was calculated from the data of Korean body size information [21], which was obtained from 1500 women and men. Bare foot means foot length from heel to ankle, and forefoot means foot length from ankle to toe; these two values were used to calculate trajectories of foot and toe joint during walking. In conclusion, ratio of toe length was 0.04 times of one`s height. In case of foot length without toe is needed to calculate the foot and toe trajectory that mentioned in next chapter.

2.2 Kinetic description of walking model

To calculate CoM trajectory, needs some constraints such as step length, walking period, step height and CoM height. Originally, with these parameters, ideal trajectory of CoM, which means that similar as the trajectory of real human in x direction can be represented as the combination of sine function as Liquan explained in [22].

All these information is achieved from the full joint kinematic model. In case of simplified walking model with ZMP, however, only the effects by movement of CoM will be observed and dealt. Movement of each joint will not be dealt seriously in this model.

In flat ground condition, ZMP can be same with center of pressure under the plantar of human and also studies about ZMP measuring of active type prosthesis [23] was done by Herr *et al*, so ZMP measuring during human walking is not so unreasonable. The concept of ZMP is able to be driven from the three dimensional inverted pendulum model [24] that described in both of Cartesian coordinate and angular coordinate. Cart table model, which from 3d inverted pendulum model is used for simplifying of walking motion. Also this model includes the information of ZMP point in x direction, which makes mass does not fall during its moving. Equation of ZMP in x direction can be written in this form

$$\vec{p}_x = \vec{x} - \frac{|\vec{z}|}{|\vec{g} + \ddot{z}|}\ddot{\vec{x}} \qquad (1)$$

Using this equation, trajectory of ZMP is able to be achieved because the trajectory of CoM is known now. Also acceleration of CoM in x direction can be calculated by the trajectory of CoM, too. This ZMP information is used as the action point of ground reaction force. The ground reaction force can be obtained by following the numerical formula as Herr mentioned in [25].

$$\vec{F}_{ground\ reaction} = m(\vec{a} - \vec{g}) \qquad (2)$$

Where m means mass of central of mass, a means the acceleration in walking direction and g means gravity. In this study, acceleration of CoM is estimated as the acceleration that in only x direction to make more simple calculation; a change of CoM in z direction is too small to

With the information of ZMP, position vector between each joint and ZMP can be calculated. Before calculation of joint torque, more position vector is needed that from the projection of the ground reaction force to the center of each joint. With this position vector, joint torque can be calculated as the production of position vector and ground reaction force. The results of this calculation is almost same as the measured data from real human walking on the force plate or calculated result through the inverse dynamics [26]. Detailed numerical formula is mentioned below.

$$\vec{\tau}_{joint} = \vec{r}_{joint\,/GRF} \times \vec{F}_{ground\ reaction} = (\vec{r}_{joint} - \vec{p}_x)\sin\theta_r \times \vec{F}_{ground\ reaction} \qquad (3)$$

θ_r means the angle between ground reaction force vector and the position vector from ZMP to the joint center. In fig. 3, this vector contribution is shown.

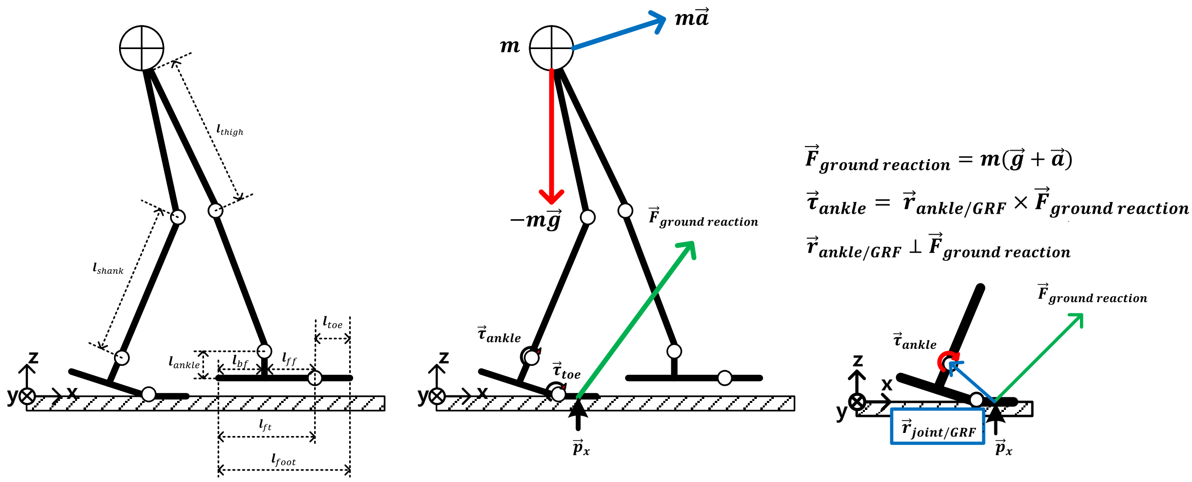

Figure 3 In the study of Jinying [4], however, no theoretical proof or modeling was exist. In this study, analysis of toe and ankle joint stiffness changing during walking by numerical modeling will be dealt. Walking model was built based on the body part ratio of Winter [19]. Dimention of reference human was 170cm of height and 63kg of weight, as the experiment of Gates [20]. Instead of inverse dynamics based full joint walking model in [22], simplified ZMP based walking model was used in this study. Left figure shows the kinematic information of suggested model; detailed information is able to be known in table 1. Middle and right figure shows the kinetic information of model; especially right figure shows how joint torque can be calculated with ground reaction force and position vector from ZMP to the center of joint. The reason that use of simplified walking model is to calculate joint torque and ground reaction force more easily than full joint inverse kinematics model. Ground reaction force of ZMP based walking model is calculated by the subtraction of inertia force and gravity. Inertia of each link will be ignored for simple calculation. The name of each segments of walking model is represented in table 1. In this simplified ZMP based walking model, joint torque can be calculated as numerical figure on the right.

2.3 Trajectory analysis of walking model

CoM trajectory of this suggested simplified walking model, however, unlike full joint kinematic model that mimics real human walking, shows smooth line. The trajectory of suggested walking model can be calculated following with some new parameters.

Trajectory of CoM, ankle joint and toe joint is used as input variable of system as mentioned before. Trajectory value followed the former study of Kajita *et al* [27]. Trajectory information was arranged again in 6 step periodic movement to make easier to divide. Toe joint trajectory, however, not mentioned in former study. So, in this study, trajectory of toe joint is specified. It was divided in 6 steps like trajectory of the foot. Time constraints are same with the trajectory of foot in [27]. Detailed toe trajectory is represented below.

$$\begin{array}{ll}
kD_s + l_{ff} & t = kT_c \\
kD_s + l_{ff} & t = kT_c + T_d \\
kD_s + L_{a0} + l_{ff} & t = kT_c + T_m \\
(k+2)D_s - l_{ab}(1 - cos q_f) + l_{ff} cos q_f & t = (k+1)T_c \\
(k+2)D_s + l_{ff} & t = (k+1)T_c + T_d \\
(k+2)D_s + l_{ff} & t = (k+1)T_c + T_m
\end{array}$$

When $t = kT_c$ and $kT_c + T_d$, x directional trajectory has same value. This means toe joint is stacked on the ground and does not move until foot kicks off the ground. Length parameter l_{ff} means the length of foot component except heel and toe, as mentioned in table 1. Z directional trajectory is justified as below.

$$
\begin{cases}
0 & t = kT_c \\
0 & t = kT_c + T_d \\
H_{a0} - l_{an} & t = kT_c + T_m \\
l_{ab} \sin q_f + l_{an} cosq_f + l_{ff} \cos q_f & t = (k+1)T_c \\
0 & t = (k+1)T_c + T_d \\
0 & t = (k+1)T_c + T_m
\end{cases}
$$

Also zero means has no potential variable during stacked on the ground. When ankle joint floats in the highest position, toe joint is positioned in lower position than ankle joint as height of ankle joint. With this position information, spline for x and z axis trajectory was drawn by MatLAB 2012b. About angle q_b and q_f will be referred in next chapter.

2.4 Joint Angle regeneration for suggested walking model

When from heel starts to take off from the ground to foot kicks the ground off, the angle between the ground and foot called as q_b. In contrast, when heel strikes the ground, the angle between foot and the ground is called as q_f. In [27], q_b angle justified without toe segment. Moreover, angle q_b and q_f in [27], maybe not suitable because these values were from low walking speed (about 0.2m/s). Therefore, in this study, q_b and q_f were newly justified with addition of toe segment.

By Jinying et al [4], toe angle during walking in 1.2m/s of speed was measured. In according to their data, the maximum angle of toe joint was 23 degree and the minimum value was 0; this data will be used for calculation of angle q_b. In case of ankle angle, recodes zero value when heel strike is occurred and -7.5 degree when foot flat is occurred right after the heel strike. This will be used for calculation of q_f. Both of ankle and toe joint, is relative angle value; ankle angle is the angle between shank and foot segment, toe joint is from the metatarsal and foot segment angle. So, when foot touches the ground or kicks the ground off, relationship of neighbored two segments should be understood. For instance, when foot kicks off the ground, toe is on the ground, so angle of metatarsal is able to keep its angle by 180 degree. However, foot segment, starts to be inclined. Therefore, toe angle, which becomes the subtraction of metatarsal and foot segment angle becomes the angle between foot segment and level ground; this becomes q_b in this study. In the same way, when heel strikes the ground, ankle angle has its own value. After heel strike, when foot flat occurs, shank does not move as much as foot, so ankle angle change is able to be described by the movement of foot segment only; this becomes the value of q_f in this study.

Through this process, both of q_b and q_f were reorganized. This value will be used when calculating the joint stiffness value with joint torque that calculated by equation (3).

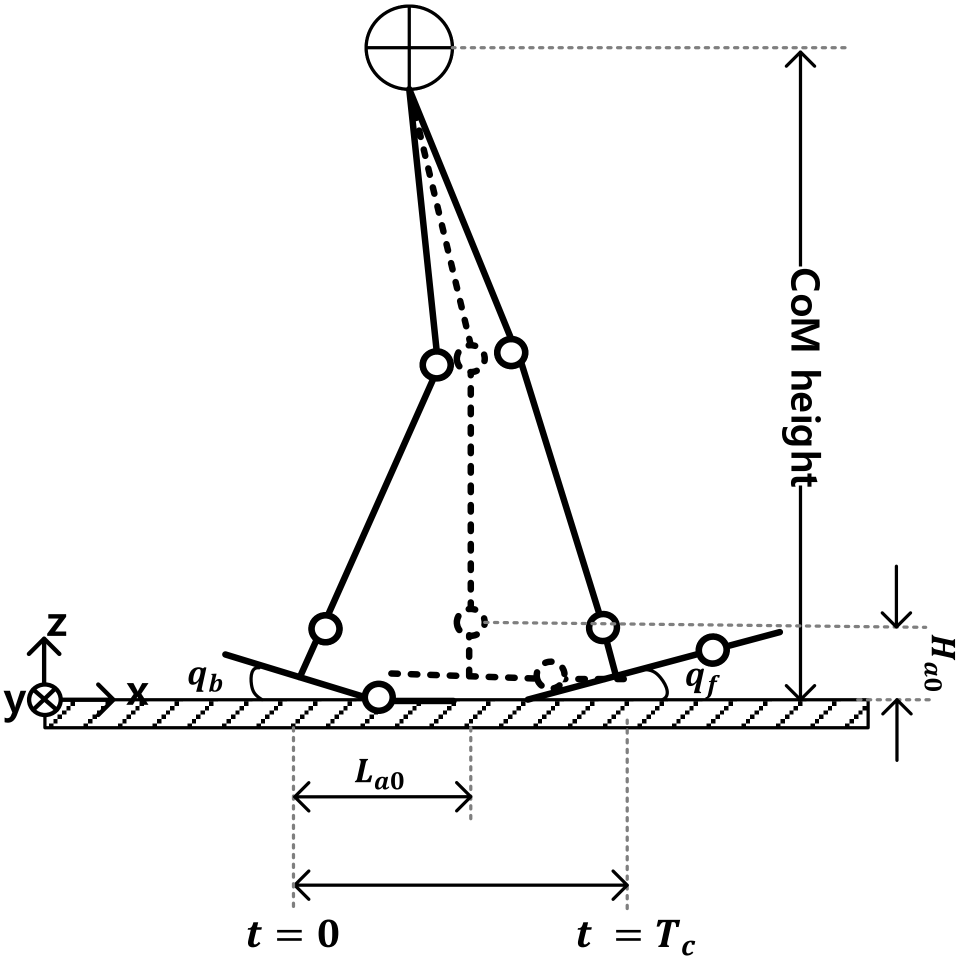

Figure 4 Setting of walking parameters and constraints. Walking period, step length, step height and height of CoM are used as parameter of calculation. Walking period and step length is defined as the time and length between the position of the rear foot and fore foot; the position of foot is defined as the relative position of each feet. In this calculation, 0.5 second for walking period, 0.14m for the highest position of the ankle joint, 0.6m for step length. CoM height was selected by the trajectory of CoM in z direction and this means the height of CoM can be changed during walking as normal people. Especially in case of step length was calculated from the category of slow walking speed (1.2m/s) and walking cadence (2step/s), which is able to be classified to slow walking speed [28]. Two angles that between ground and foot are also used to calculate the trajectory of ankle and toe joints. In this research, the angle qb and qf was newly justified because of change of walking speed and addition of the toe joint.

2.5 Numerical analysis

Before build the walking model, numerical analysis with block diagram is needed. In the first block diagram that represented in Figure 3-c, shows the relationship between acceleration of CoM of body and joint torques. Also this diagram includes terms of ankle and toe joint trajectory, so joint torque can be calculated with information of position vector between joint and ZMP point. Trajectory of ZMP is able to be achieved with the information of CoM trajectory and acceleration. As mentioned in equation (4), with these information, joint torque can be achieved. All these things were mentioned before and represented in block diagram that referred in Figure 3-c. In Figure 3-d, block diagram of stiffness calculation is represented. In this block diagram, reverse calculation of joint angle through joint torque is done. However, joint torque; the input of this block diagram, joint angle; the output of this block diagram are only known. The plant of this system is unknown. So, in this block diagram, system identification should be processed to know the stiffness of joints. Detailed contents of system identification is written in appendix. By this results, angle-torque curve of each joint of walking model can be drawn. Then, the stiffness of each joint can be evaluated that suitable for natural walking like normal human.

2.6 Walking model simulation

To simulate the walking simulation, Matlab™ 2012b was used. In this part, contents of programming will be referred simply. Total programing code is written in appendix.

Unlike point feet walking, segmented feet, which has additional toe joint needs rotation of foot and toe. Especially, toe need to move in two phase; cannot be rotated from heel strike to heel off during stance phase. For this movement, boundary condition for foot movement was added; (in writing). This movement was realized by SimMechanics block diagram. Ground reaction force, which occurring during stance phase, was separated in 2 ways. In detail, when one leg is on the ground, only one ground reaction force is acting on the foot. However, during double support phase, both of right and left foot is affected by ground reaction force. With simplified equation of ground reaction force that mentioned in equation (2), ground reaction force is programmed that can be acted on the both leg when they are on the ground. Finally, change of joint stiffness was one of the challenge of this study. Joint stiffness change condition has different condition to be processed; stiffness change timing for ankle joint and toe joint was that. Based on stiffness information that will be mentioned in next chapter.

2.7 Experiment setup

Joint stiffness is approximately achieved by the angle-moment curve of joint during walking. By rotational movement of joint during walking, joint angle is defined as the relative angle of adjacent two segments. Ankle angle is defined as the angle of foot with respect to the shank with 90 degree of additional angle because foot is linked with shank segment in rectangular, unlike other segments that linked directly to each other. Toe joint is defined as the angle of foot with respect to the angle of metatarsal. By these relative rotational movement and joint torque, joint stiffness can be defined as like this formula below.

$$\mathrm{K}_{joint} = \frac{\mathrm{d}\vec{\tau}_{joint}}{\mathrm{d}\vec{\theta}_{joint}} \qquad (4)$$

By equation (4), joint torque can be calculated again with terms of stiffness K and joint angle theta

To evaluate the walking trajectory of walking model, CoM trajectory should be obtained. Before considering about the CoM trajectory of suggested walking model, ideal trajectory of CoM that similar as real human barycenter trajectory during walking should be dealt to use as reference. To calculate the ideal trajectory of center of mass of human during walking in x direction, a simple sine function is needed that mentioned in [22].

With stiffness calculation that mentioned before, joint stiffness can be calculated. In the data of angle-moment curve of slow walking speed (1.2m/s) [28], joint stiffness will be calculated and the result will come out in the form of the slope of curves; walking speed in 1.2m/s can be classified as slow walking speed, however, 1.2m/s is the boundary of slow walking. By Jinying *et al*, it can be represented as “comfortable walking speed” [4], so this value is able to be treated as walking speed in ordinary life. Moment angle curve was from the data of Jinying in [4]. Linearized joint stiffness is displayed in figure 4. These linearized value will be adapted for joint torque calculation that calculated from block diagram that mentioned in earlier chapter. Some assumption are existing; first, regardless of joint stiffness, joint angle change during walking is same as before stiffness change; to make more easier torque calculation. Also, CoM height follows the z directional trajectory; to make easier inverse calculation of joint torque that includes the 2^{nd} differential terms of x and z directional trajectories. Finally, inertia of all segments were ignored for simplified calculation.

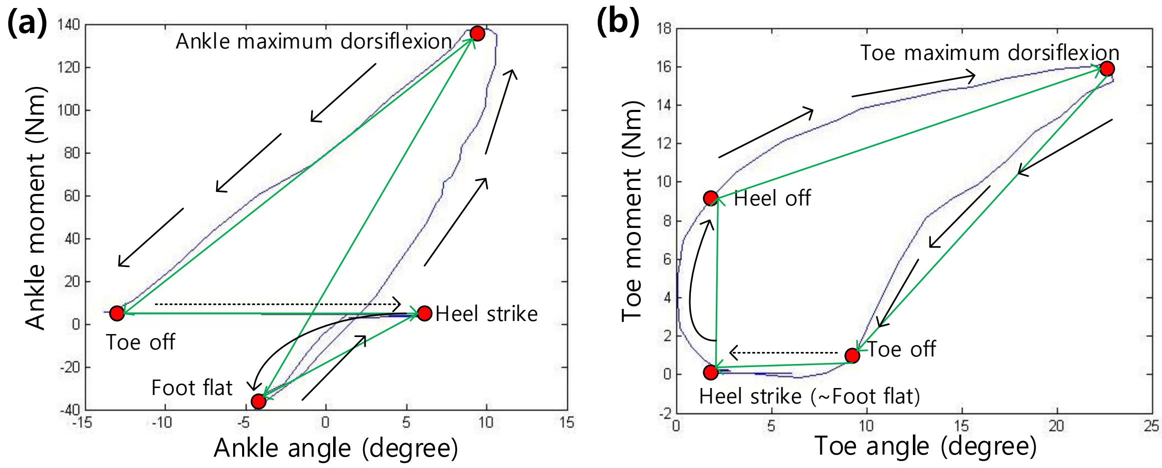

Figure 5 Angle moment curve at the (a) ankle joint and (b) toe joint that measured in [4]. Arrows means the direction of the curve; means the walking cycle progression. All of these curves were drawn again in Matlab 2012b. Value of joint stiffness were selected by the tilt of representative two points of each curve. In (a), position of heel strike, foot flat, ankle maximum dorsiflexion and toe off were selected as the position that makes linear line. In (b), also, heel strike, toe off position were selected. Instead of foot flat and ankle maximum dorsiflexion position, in (b), heel off, toe maximum dorsiflexion point were selected. With linear lines that links two different point, joint stiffness can be calculated as the tilt of linear line as mentioned before. Between toe off and heel strike region, joint stiffness was reffered as 0 because in this phase, foot does not touches the ground; swing phase. Calculated value of joint stiffness will be used for change the joint torque value that can be represented as terms of joint angle and joint stiffness; assume that joint angle is same in all cases. Then, joint torque that can be represented as terms of position vector and ground reaction force that consists with x direacional acceleration. In summary, change of joint stiffness, can cause the change of x directional acceleration of body, so this can affect to the whole body movement during walking.

Ⅲ. Results

3.1 Trajectory calculation results

3.1.1 CoM trajectory

All kinds of trajectory was generated by MatLAB 2012b. First, trajectory of CoM, which means the movement of whole body was generated. Both of X directional and Z directional trajectories are shown in Figure 6-a, b. As mentioned before, X directional trajectory shows like combination of linear and sinusoidal function. In contrast, Z directional trajectory shows almost like sinusoidal function. These two data means the movement of human body during walking shows 1) repeats fast and slow moving through X direction (walking direction) and 2) goes up and down during walking in Z direction (vertical direction). Also, step length, which was defined as 0.6m in previous chapter can be shown in Figure 7-a. With 0.5s of terms of step cycle, X directional distance increases 0.6m per one step cycle.

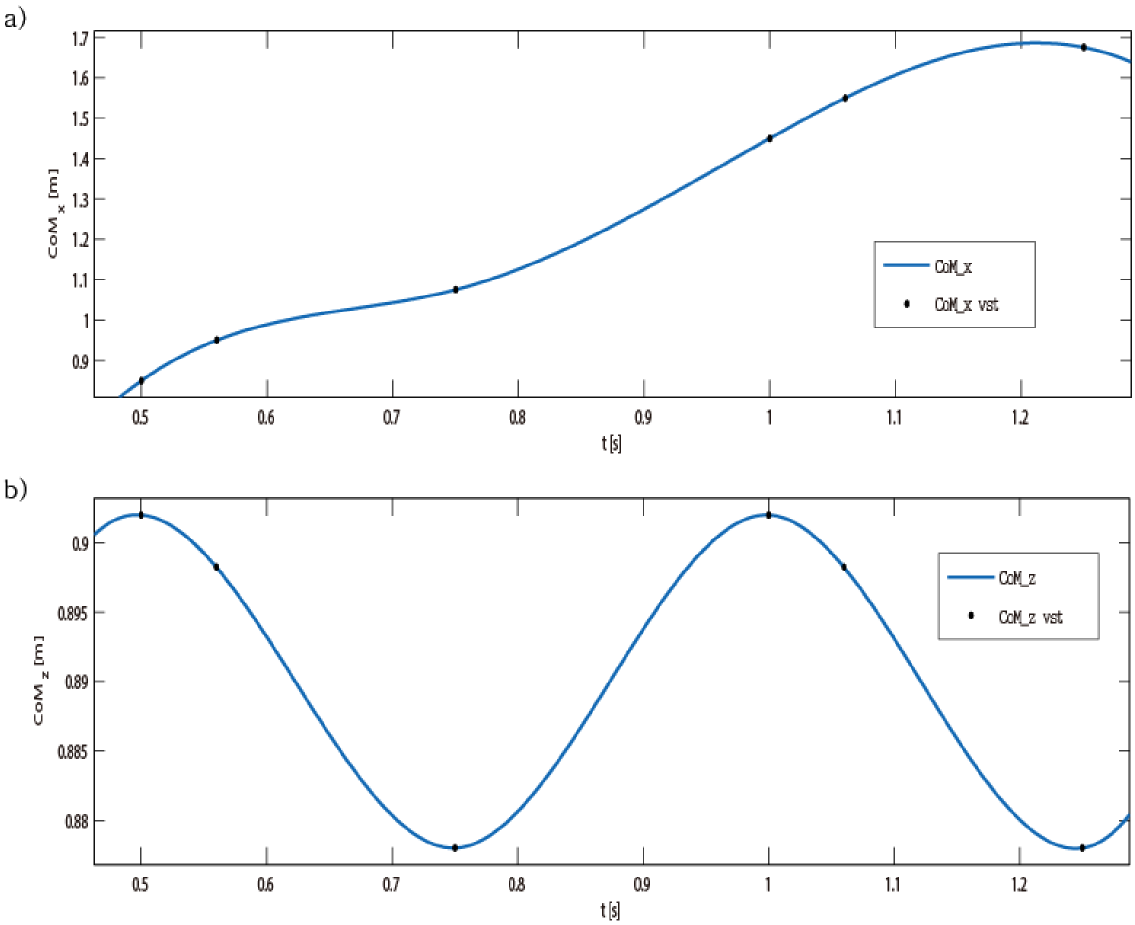

Figure 6 a) X directional and b) Z directional trajectory of CoM. In case of X directional movement, trajectory shows the combination of linear and sinusoidal function. It repeats two times during one whole step cycle, so this means CoM moves by movement of lower limb through walking direction. In contrast, in case of Z directional trajectory shows only sinusoidal function form. This means body repeats goes down and rises up during walking by movement of lower limb. As mentioned in chapter 2, this changes are mainly decided by walking parameters, such as step length, height of CoM and the maximum length between CoM and each feet.

3.1.2 Ankle joint trajectory

Ankle joint trajectory is shown in Figure 7. Also, both of X and Z directional graph are all generated. When comparison with [27], except its dimension, ankle trajectory shows similar form as [27]. This trajectory is decided by body dimension that mentioned in Table 1 and contact angle with the ground. However, as mentioned in chapter 2, by addition of toe, angle q_b was increased than the value that suggested in [27]. Therefore, the moment about toe off, ankle joint trajectory, especially X directional trajectory shows different shape in contrast to [27].

3.1.3 Toe joint trajectory

Toe joint trajectory is originally generated in this study and shown in Figure 8. Trajectory of toe joint shows similar form as ankle joint trajectory as mentioned in previous section because toe joint is attached to foot segment, which is connected to the ankle joint. Additionally, both of X and Z directional trajectory shows little bit of difference with ankle joint trajectory because of position difference. However, zero position in Z direction means toe does not kick the ground off and still attached on the ground. This makes little bit of difference between Z directional ankle joint trajectory.

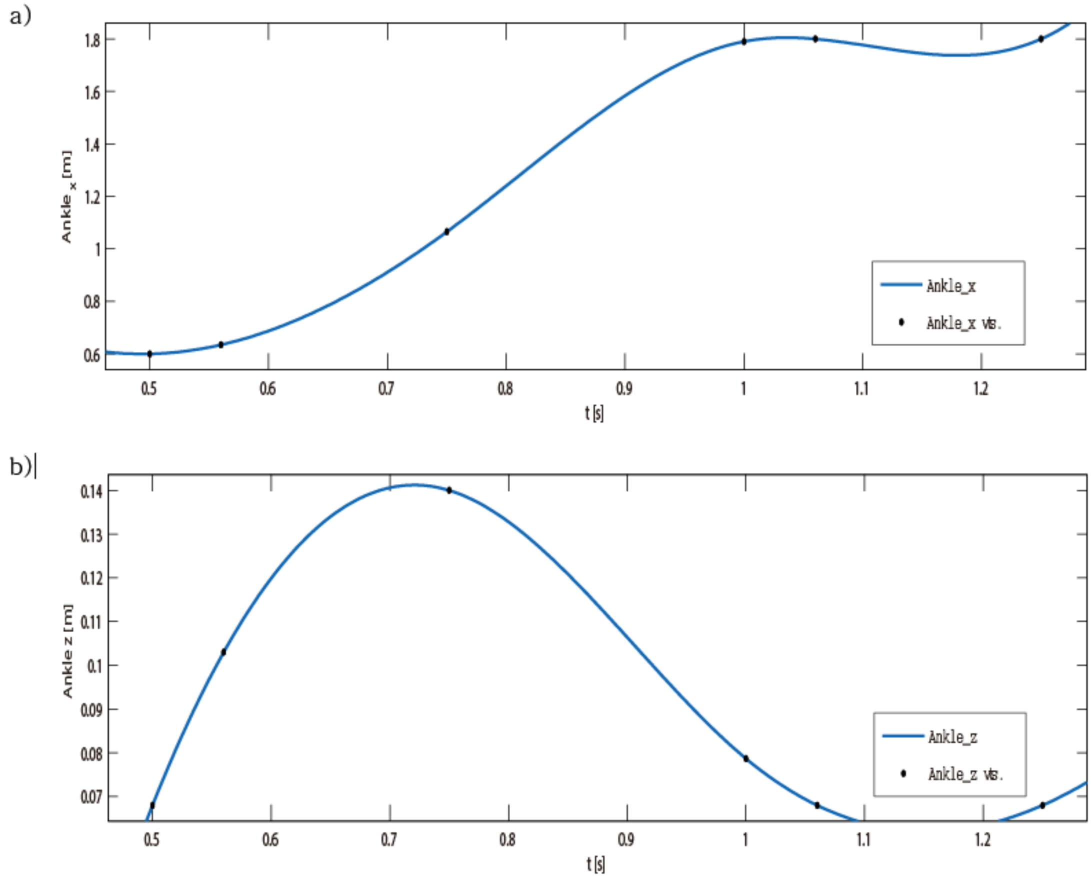

Figure 7 a) X directional and b) Z directional trajectory of ankle joint. Because of change in walking speed, the angle q_f and q_b was changed as mentioned in chapter 2. Unlike trajectory of CoM, this trajectory is mainly decided by several body dimensions such as height of ankle, length of forefoot. Angle q_b and q_f effects when foot try to strike the ground with its heel and kicks the ground off with its toe. In this study, by addition of toe, especially q_b that means the angle between the ground and the feet that kicks the ground has been changed largely (4 times enlarged than previous study). Therefore, when feet tries to kick the ground off (when t= kT_c+T_d), ankle joint can have longer position in contrast to previous study by its trigonometric function term.

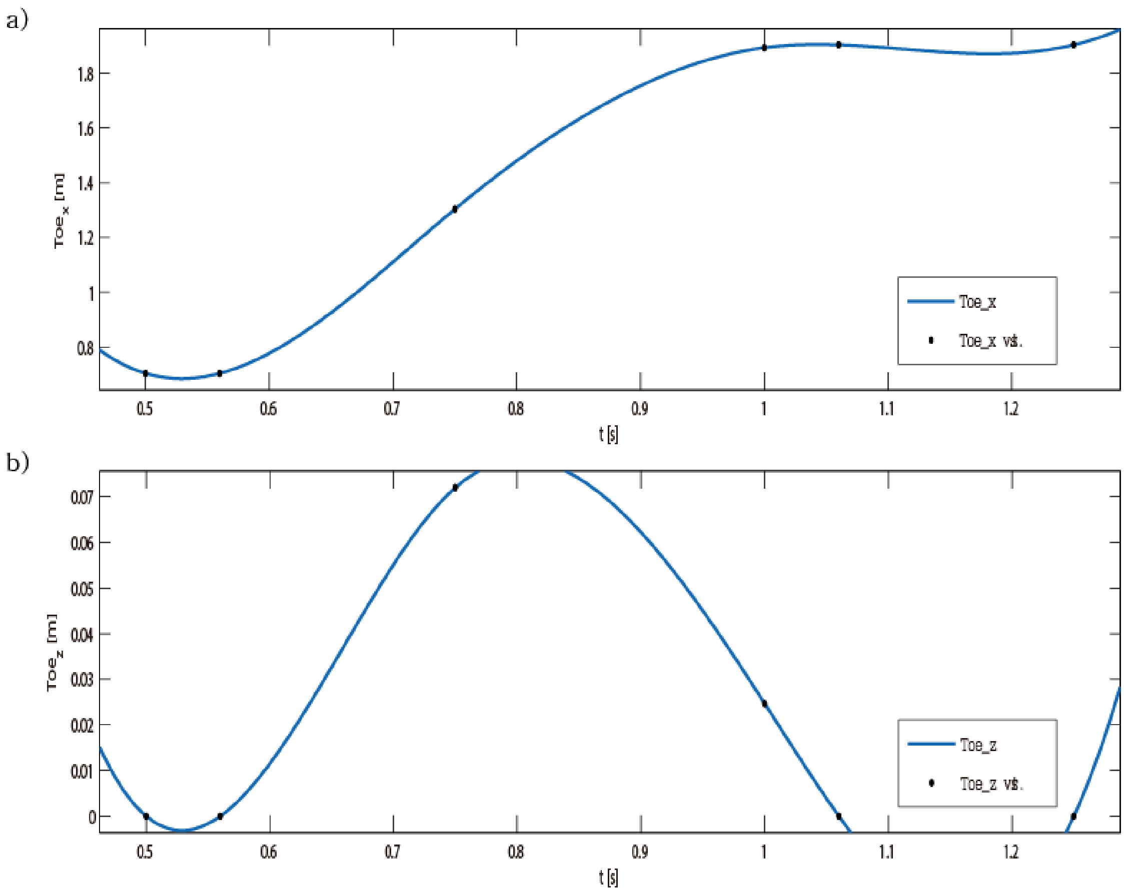

Figure 8 a) X directional and b) Z directional trajectory of toe joint. This trajectory is added in this study originally, unlike previous study that did not considered about toe joint. Originally, toe joint is attached to the foot segment that linked to the ankle joint. Therefore, in case of X directional trajectory of toe joint has similar form with ankle joint. Toe joint is positioned in precedent position than ankle joint, so X directional trajectory of toe joint should have larger value at the same time than ankle joint trajectory. In case of Z directional trajectory, toe joint should keep its position on the ground before toe off of walking cycle. Therefore, position of Z directional toe joint keeps zero value until toe off is occurred (when $\mathrm{t} = \mathrm{k}T_c + T_d$). When feet touches the ground and foot flat is occurred again, toe joint has zero value again.

3.2 Ground reaction force calculation

Movement of CoM is divided in two ways; X and Z direction, as mentioned before. Therefore, ground reaction force should be calculated in both of two directions. Both of two values will be used to calculate joint torque during walking.

3.2.1 X directional ground reaction force

Unlike Z directional ground reaction force, X directional ground reaction force not includes terms of gravity and only includes x directional acceleration. Therefore, this value is described as the force that from the acceleration of body movement. This is shown in Figure 9-b. However, it shows very large amount that close in upon 70% of body weight. Originally, this value should not exceed maximum 50% of body weight when walking [30]; notice that walking speed that chosen in [30] is 1.5m/s, which is faster than the value that chosen in this study, so ground reaction force in [30] can be larger than the theoretical value in this study. This problem will be described in discussion chapter.

3.2.2 Z directional ground reaction force

Z directional ground reaction force, which is decided by body mass and z directional acceleration, as mentioned in chapter 2 is calculated and shown in Figure 9-a. It appears as one of sinusoidal function that repeats increasing and decreasing. This is quite different form when compared to the real human ground reaction force measurement result that starts from zero and ends to zero again and this is described in many biomechanics studies [29]. Fortunately, the value of Z directional ground reaction force is not so different from theoretical value, which is 110% of body weight at the maximum value and 80% of body weight when the minimum value [29].

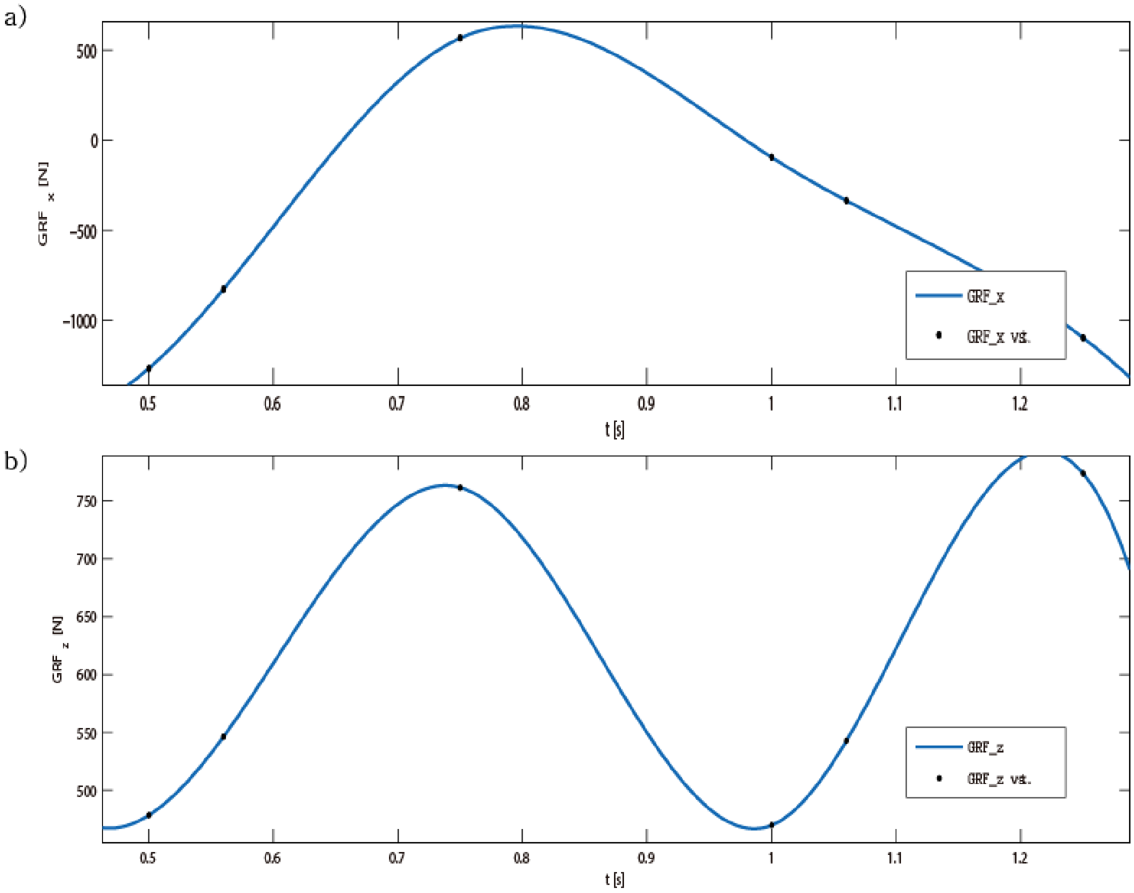

Figure 9 a) Horizontal (X direction) and b) vertical (Z directional) ground reaction force. Both of ground reaction force are able to be obtained by body mass, gravitational acceleration (in case of vertical force) and acceleration for each direction. In case of horizontal ground reaction force, however, shows larger value (70% of body mass) than theoretical value (20% of body mass). In case of vertical ground reaction force, has a sinusoidal form and 120% of body mass for the maximum value, 80% of body mass for the minimum value. Although it has no zero value, its sinusoidal form means the whole walking cycle is done.

3.3 ZMP trajectory

As mentioned before, ZMP trajectory is used to calculate joint torques that represented as vector production as mentioned in chapter 2. Originally, this should be calculated both of X and Y direction, however, in this study, only sagittal plane will be described, so ZMP through X direction is the only one consideration. This term is represented as x directional position and z directional acceleration and gravitational acceleration. Therefore, trajectory of ZMP should show similar form as X directional CoM trajectory. Result is shown in Figure 10. Unlike [27], in Figure 12, only ZMP trajectory is shown, without safety region. This is because x_ed and x_sd is given as its maximum value, not following the algorithm that mentioned in [27].

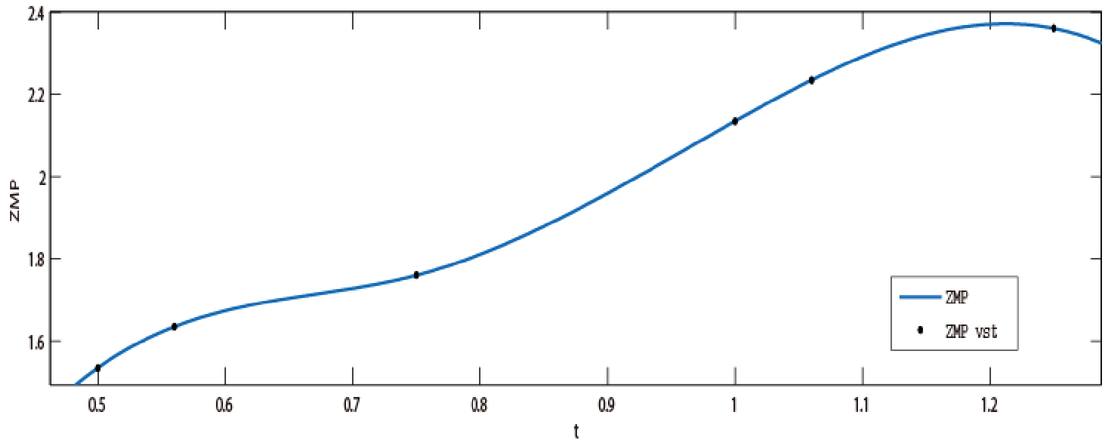

Figure 10 Trajectory of ZMP in X direction. ZMP is able to be treated as the center of pressure under the plantar as mentioned in chapter 2. Therefore, this trajectory keeps its position under the plantar and has zero value for Z direction. Moreover, in this study, only movement in sagittal plane is in consideration, so only X directional ZMP trajectory is shown. By its theoretical formula, ZMP tracks X directional trajectory of CoM. This figure shows suddenly increasing form. When comparing this figure with figure 6-a, it shows similar form with trajectory of CoM in X direction. This means the movement of center of pressure under the plantar by movement of body mass during walking.

Ⅳ. Discussion

Calculation of ground reaction force through body mass and movement of CoM was done. Although the value of calculation was not so reasonable, however, in case of Z directional ground reaction force shows completely one walking pattern with its sinusoidal pattern. Therefore, this data is able to be used as ground reaction force data. In case of X directional ground reaction force, however, shows largely different value than theoretical data. This is caused by the small value of walking cycle term. By this phenomenon, calculation of joint torque and joint stiffness was meaningless. To solve this problem, reorganization of CoM trajectory in X direction should be considered. Direct detection of human body acceleration through motion sensor can be one of the solution. Walking speed can be controlled constantly with treadmill. Also, trajectory of each joint can be more easily obtained by this method. Moreover, the method that mentioned in this study [27] is for mainly humanoid study, so use of data from real human walking can be more accurate than the data from [27]. Joint angle detection can be obtained by motion sensor during human walking and this will be more accurate that joint angle calculation through joint trajectory. Firstly, joint angle calculation was planned to calculate through SimMechanics tool of MatLAB 2012b, however, block diagram of SimMechanics cannot be realized in this study. Moreover, in written MatLAB code, no contents that describes the periodic movement of each feet, for instance, touches the ground, double support phase or kicks the ground off was exist. By this problem, animation of walking model could not be realized, so the minimum movement of walking model. The only one cue that we can discuss about these trajectories is comparing with the data of [27]. Both of CoM and ankle joint trajectory shows similar graph form. However, comparison in numerical dimension is not reasonable because in [27], walking parameters such as walking speed, step length, double support time are different with this study. This is one of the biggest problem of this study, so in next study, realization of walking model simulation based on joint trajectory will be major work.

During feet is on the ground, Z directional position of toe joint should keep zero value as mentioned in previous chapter. However, in figure 8-b, graph shows non-zero section during feet on the ground, especially after heel strike. This problem is caused by the limitation of spline of MatLAB 2012b. Although value for Z directional position was set to zero at the specific points, there is no fixed value between two zero points. Moreover, this graph was drawn by cubic spline fitting by curve fitting tool of MatLAB 2012b. With this tool, drawing cubic spline that links several points is available, however, cannot keep its value at fixed section by user. This problem causes not correct form of trajectory. To solve this problem, spline that has more specified points can be used. As mentioned before, use of the trajectory function from real human walking data that obtained by clinical test with three dimensional motion sensor also can be help to solve this trajectory based simulation problem.

Ⅴ. Conclusion

Through this study, ground reaction force calculation through given CoM trajectory was done. Also, joint trajectory was calculated to calculate ZMP position. Originally, the effect of ankle and toe joint stiffness to the whole body movement during walking should be observed, however, as mentioned before, due to inaccurate result of X directional acceleration, joint torque cannot be calculated. This was the biggest limitation of this study. Except this problem, spline based joint trajectory generation was treated as limitation because of its inaccuracy. By these limitations, unfortunately, this should be postponed to the next studies because of problems that mentioned in last chapter.

To make this study higher completeness, largely two challenges will be dealt. First, based on real human motion tracking test, continuous function of each joint and body will be obtained. As mentioned before, with this method, accurate trajectories for each joint can be obtained, not based on the splining of specified points. Second, additionally, acceleration of human body during walking should be obtained to calculate more accurate joint torque than trajectory based method that used in this study. Joint stiffness is based on the joint torque value as mentioned in chapter 2, therefore, accurate joint stiffness calculation is essential for this study. Use of three dimensional motion sensor system as mentioned before, also can be help of calculation of joint stiffness value by obtaining of joint angular velocity and acceleration; use of this method can be more complicate than ground reaction force based torque calculation, but more accurate calculation can be available.

After finish all these works, this study will be applied to several fields. One of example is designing the orthosis that equipped variable stiffness function for drop-foot patients that has trouble to walk like normal people. Based on the data from the seriated study, joint stiffness adaptable orthosis can vary its own stiffness value for natural walking of drop-foot patient. Then, joints of patients can have its suitable value for walking and they can walk as healthy normal people.

Appendices

Appendix 1 - MatLAB programming code (m-file)

Trajectory calculation of center of mass and time normalization to walking cycle

```
D_s = 0.6;
x_ed = 0.25;
x_sd = 0.25;
T_c = 0.5;
T_d = 0.06;
T_m = 0.25;
for k = 1
    t = [k*T_c k*T_c+T_d k*T_c+T_m (k+1)*T_c (k+1)*T_c+T_d
(k+1)*T_c+T_m];
    t_CoM(1,3*(k-1)+1:3*k+3)=t;
    x = [k*D_s+x_ed (k+1)*D_s-x_sd (k+1)*D_s-0.5*x_sd (k+1)*D_s+x_ed
(k+2)*D_s-x_sd (k+2)*D_s-0.5*x_sd];
    z = -0.012*sin(4*pi*(t-2.62))+0.89;
    x_CoM(1,3*(k-1)+1:3*k+3) = x;
    z_CoM(1,3*(k-1)+1:3*k+3) = z;
end
t_nor_cyc = (t_CoM-T_c)/(k*T_c+T_m);
% Execute cftool and save the loaded curve (t_CoM vs x_CoM and
z_CoM)
% to the workspace with name x_CoM and z_CoM
% After that, use [@t @tt] = differentiate(@_CoM,t_CoM) to get the
2nd
% differentiate of trajectory, then calculate the p_x
```

Trajectory calculation of ankle and toe joint

```
q_b = 0.4; %[radian]
q_f = 0.13; %[radian]
L_a0 = 0.4654;
H_a0 = 0.14;
l_an = 0.068;
l_bf = 0.0867;
l_ff = 0.1717;
l_ft = 0.1037;
t_foot = zeros(1,6);
x_foot = zeros(1,6);
z_foot = zeros(1,6);
t_toe = zeros(1,6);
x_toe = zeros(1,6);
z_toe = zeros(1,6);
for k = 1
    t = [k*T_c k*T_c+T_d k*T_c+T_m (k+1)*T_c (k+1)*T_c+T_d
(k+1)*T_c+T_m];
    t_foot(1,3*(k-1)+1:3*k+3)=t;
    x_a = [k*D_s k*D_s+l_an*sin(q_b)+l_ft*(1-cos(q_b)) k*D_s+L_a0
(k+2)*D_s-l_an*sin(q_f)-l_bf*(1-cos(q_f)) (k+2)*D_s (k+2)*D_s];
    x_foot_mod(1,3*(k-1)+1:3*k+3)=x_a;
    z_a = [l_an l_ft*sin(q_b)+l_an*cos(q_b) H_a0
l_bf*sin(q_f)+l_an*cos(q_f) l_an l_an];
    z_foot_mod(1,3*(k-1)+1:3*k+3)=z_a;
end
for k = 1
    t = [k*T_c k*T_c+T_d k*T_c+T_m (k+1)*T_c (k+1)*T_c+T_d
(k+1)*T_c+T_m];
    t_toe(1,3*(k-1)+1:3*k+3)=t;
    x_t = [k*D_s+l_ft k*D_s+l_ft k*D_s+L_a0+l_ft (k+2)*D_s-
```

```
l_an*sin(q_f)-l_bf*(1-cos(q_f))+l_ft*cos(q_f) (k+2)*D_s+l_ft
(k+2)*D_s+l_ft];
    x_toe_mod(1,3*(k-1)+1:3*k+3)=x_t;
    z_t = [0 0 H_a0-l_an (l_bf+l_ft)*sin(q_f) 0 0];
    z_toe_mod(1,3*(k-1)+1:3*k+3)=z_t;
end
```

ZMP, GRF and position vector from GRF to center of joint calculation code

```
m = 63; % mass
g = 9.8; % gravity acc.
p_x = x_CoM_s-((z_CoM_s.*transpose(xtt))/(transpose(ztt)+g)); % ZMP
trajectory
grf_vert_mod = m*(g+transpose(ztt));
grf_horz_mod = m*transpose(xtt);
r_foot_mod = transpose([x_foot_mod; zeros(1,6); z_foot_mod]);
r_toe_mod = transpose([x_toe_mod; zeros(1,6); z_toe_mod]);
p_x_vec_mod = transpose([p_x; zeros(2,6)]);
r_zmp_foot_mod = r_foot_mod-p_x_vec;
r_zmp_toe_mod = r_toe_mod-p_x_vec;
grf_vec_mod = transpose([grf_horz; zeros(1,6); grf_vert;]);
angle_ankle_mod = atan2(z_foot_mod,x_foot_mod-p_x);
angle_toe_mod = atan2(z_toe_mod,x_toe_mod-p_x);
angle_grf_mod = atan2(grf_vert_mod,grf_horz_mod);
angle_ankle_vec_mod = (pi/2)-angle_ankle_mod;
angle_toe_vec_mod = (pi/2)-angle_toe_mod;
angle_grf_vec_mod = (pi/2)-angle_grf_mod;
angle_torque_ankle_mod = [angle_ankle_vec_mod+angle_grf_vec_mod;
ones(2,6);];
angle_torque_toe_mod = [angle_toe_vec_mod+angle_grf_vec_mod;
ones(2,6);];
r_torque_ankle_mod =
sin(angle_torque_ankle_mod).*transpose(r_foot_mod-p_x_vec_mod);
r_torque_toe_mod = sin(angle_torque_toe_mod).*transpose(r_toe_mod-
p_x_vec_mod);
torque_ankle_mod = cross(transpose(r_torque_ankle_mod),grf_vec_mod);
torque_toe_mod = cross(transpose(r_torque_toe_mod),grf_vec_mod);
```

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요 약 문

ZMP 기반의 이족 보행 모델을 이용한 관절 강성 변화로 인한 보행 변화의 수학적 해석

본 논문은 보다 사람에 가까운 보행 조건을 찾기 위하여 발목과 발가락 관절의 강성을 보행이 진행되는 동안 그 값을 변화시킴으로써 보행에 미치는 영향을 찾아내고, 또한 사람의 보행에 가까운 결과를 얻기 위하여는 어떤 강성 값을 이용해야 하는지에 대하여 연구하였다. 보행이 진행되는 동안, 실제 사람의 다리에서는 근육 및 인대의 영향으로 인하여 하지의 각 관절의 강성 값이 상황에 맞게 변화하는 특성을 지니고 있다. 이에 해당 연구에서는 간단한 ZMP 기반의 이족 보행 모델을 이용하여 보행에 미치는 관절 강성의 영향에 대하여 다루고자 한다. 관절에서 생성되는 회전력은 지면 반력과 지면 반력으로부터 수직으로 관절의 중심까지 그은 위치 벡터와의 곱으로 그 크기를 추산할 수 있으며, 이는 다시 관절이 이루는 각도와 강성의 관계로 나타낼 수 있다. 하지 관절 각도가 변하지 않는다는 가정 하에, 강성을 변화시켜 그로 인한 신체 전체의 움직임의 변화를 관찰하였다.

핵심어: ZMP 보행 모델, 관절 회전력, 지면 반력, 관절 강성