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Muscle Force Analysis of Human Foot Based on Wearable Sensors and EMG Method

Enguo Cao, Yoshio Inoue, Tao Liu and Kyoko Shibata
Kochi University of Technology,
Japan

1. Introduction

Sports injuries, traffic accident, childhood diseases and life-style related diseases may cause movement disorder, and especially in aged persons, physical deterioration may leads to high risk of degeneration of motor function and falling in movement problems. Nowadays, in the trend of population aging of many countries, a large proportion of movement disorder patients are difficult to recover completely in the recent medical condition. Therefore, muscle strength and motor function recovery have aroused general concern from the society, musculoskeletal analysis and rehabilitation activities have been greatly progressed in many countries (Yoon et al., 2010), (Vallery et al., 2007), (Peshkin et al., 2005). Furthermore, modern robots are on high speed developing, but multiple-units foot has seldom caught scientists' attention, robot feet are generally treated as a single rigid part which makes gait seems clumsiness (Park et al., 2009), (Ishida et al., 2003). The human walking style is adequate for various situations compared with wheeled mobile robot, and understanding the muscle works of human limb is implemental for medical diagnosis. It is valuable to develop a multiple-units foot for flexible walking robots and rehabilitation training robots. With economical considerations, developing an easy-operating muscle diagnostic system of multiple-units foot is practical for clinical applications.

Muscle force estimation is not easy because some muscles are locating at inner part of limbs and human foot is a complex musculoskeletal system. For direct measurement of muscle forces, electromyography (EMG) method has been frequently applied as a standard clinical tool in identifying activation level of muscles, but in the EMG method the muscle activity result is relative large or small with the unit of % rather than quantitative values with the unit of Newton (Lee et al., 2008), (Merlo et al., 2003). Moreover through surface EMG method only activities of surface muscles could be estimated, but through needle EMG method the invasiveness of needles may cause reluctance of patients (Kizuka et al., 2009). For indirect estimation method, the quantitative muscle forces of human foot could be calculated, but acquirements of motions of limbs, forces applied on limbs, and the construction of musculoskeletal model are essential (Hou et al., 2007).

Up to the present there are various methods to measure human motion and ground reaction force (GRF) for limb dynamics analysis. But many of them are restricted to laboratory environment and sick to adapt to different situations. The commercial motion camera system, which is regarded as the most popular instrument and a standard tool in measuring movement of human limb, could be performed only in laboratory environment and

expensive for implements (Ferreira et al., 2009). The force plate, which is the widely used in measuring GRF with high accuracy in various fields, is limited in single stride measurement (Moustakidis et al., 2008). In this situation, wearable sensor systems based on miniature type force sensors, acceleration sensitive units and gyroscopes play more important role in the applicability for continuously walking and climbing stairs or slopes (Liu et al., 2009), (Boonstra et al., 2006). For construction of musculoskeletal dynamic model, the AnyBody Modeling System, which works in a minimum fatigue criterion way, could be introduced in various applications of variable situation. The geometric data of the model could be decided based on measurements of human anatomy model, and through inverse dynamics method quantitative results of muscle forces of human foot could be calculated. Furthermore, the EMG system could be introduced to validate the calculated muscle force results.

In this chapter, muscle forces of human foot were estimated through inverse dynamics method in AnyBody Modeling System based on wearable sensors and EMG method. A wearable sensor system was developed to measure rotational angles, GRF and centres of pressure (COP) of human limbs. Moreover one group of measured GRF and COP data are transferred into two group of GRF on two loading points in the musculoskeletal model. In AnyBody Modeling System a model of multiple-units foot was established to calculate quantitative muscle forces by inverse dynamics analysis. The shank was also built in the model for attaching muscles between foot and shank, and for the angular acceleration of the shank has influence on these muscles in inverse dynamics method. To make the model suit to different individuals, the size factor was indicated based on the length of human foot. To validate our estimation results of muscle forces, a personal surface EMG system was adopted in the experiment and the muscle activation level of Anterior Tibialis (ANT TIB), Peroneus Longus (PL), Gastrocnemius (GAST) and Soleus (SOL) were directly measured. The raw EMG result measured by hard type sensor system was filtered and rectified into integral EMG results, and the integral EMG results were introduced in the validation with the muscle force results of AnyBody Modeling System.

2. Methods

2.1 Wearable sensor system for measuring angular motion, GRF and COP

Angular motions of shank and thigh were measured by two wearable sensors with acceleration sensitive units and gyroscopes. And angular motions, GRF and COP of multiple-units foot were measured by a pair of developed instrument shoes with both force sensors and motion sensors. The measured data were expressed in a general coordinate system which was aligned with the orientation of the shoe, and located on the contacting plane between the shoe and the floor. As shown in Fig. 1, the Z-axis was made vertical, and the *Y*-axis was chosen to represent the anterior-posterior direction of the shoe on the interface plane contacting with the ground, while the X-axis was chosen such that the global coordinate system would be right-handed (Liu et al., 2008). Accurate location of the sensors is not necessary because the limb segments are regarded as rigid bodies in the study.

In this way the dispersion force on the bottom of shoe was expressed by concentrated GRF and COP, and the measured electrical data of angular motion, GRF and COP of multiple-units foot were transmit into a signal process box, which took the responsibility for wave filtering, data consolidation and transmission, as shown in Fig. 2. The signal process box also played a role of signal synchronization by sending a start pulse to the wearable sensor system combined on the leg, as all the sensors have the same sampling frequency. The
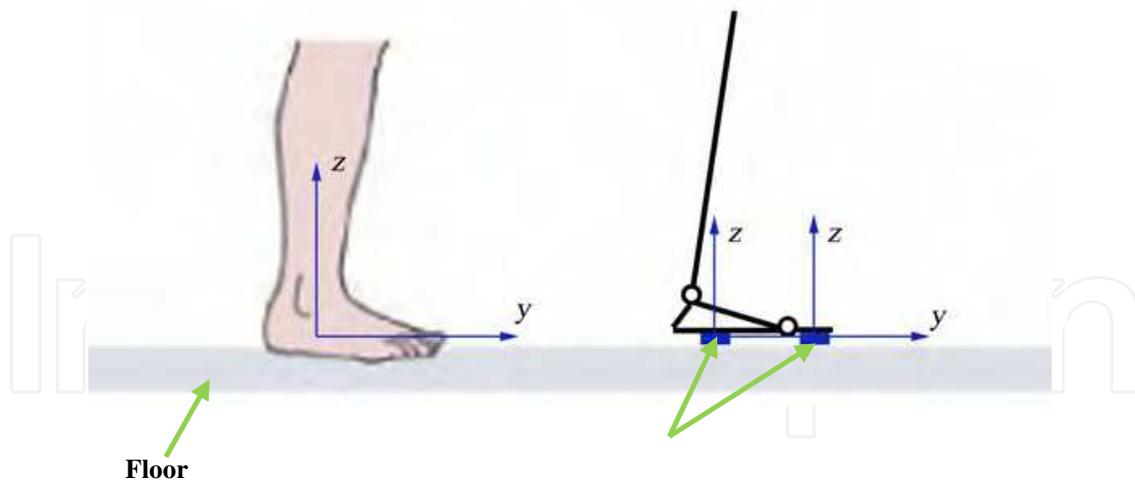
Fig. 1. Ordinate system of an instrument shoe with multiple sensors to obtain GRF, COP and angular motion of multiple-units foot.
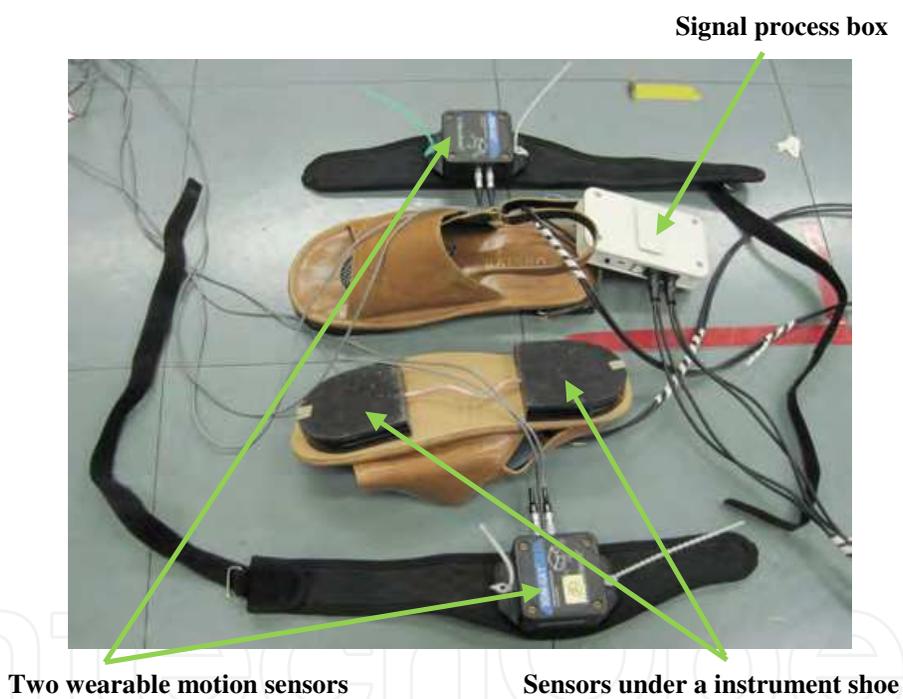

Fig. 2. The instrument shoes and two wearable sensors designed to obtain GRF, COP and angular motions of lower limbs during successive walking trial.

motion of shank was measured because muscles joint points in shank have non-separable relationship with the muscles driving the foot, and the process of gait is relied on the co-operation of these muscles. The angular motion of thigh was not involved in the calculation of muscle forces of the foot because there are no muscles connect from thigh to the foot directly, although the motion of thigh was measured in this experiment for the future research. In the wearable motion and GRF sensor system, all the sensors had the sampling frequency of 100 Hz, and the GRF, COP, multiple-units foot and shank rotations had the units of Newton (N), millimeter (mm) and degree (°) respectively. Totally four groups of

GRF, two group of COP and four groups of angular of multiple-units foot were measured by the instrument shoes, moreover one groups of angular of shank and one groups of angular of thigh were measured by the wearable sensors.

2.2 AnyBody modeling system for establishing multiple-units foot model

In human body muscles are activated by the central nervous system (CNS) base on a complicated electro-chemical process. Determining the activation that realizes a desired movement requires an extremely intricate control algorithm. The AnyBody Modeling System is not only a professional musculoskeletal modeling system, but also a kinematics and kinetics analysis system, in which inverse dynamics method is adapted to quantitatively estimate muscle forces. AnyBody imitates the workings of the CNS by computing backwards from the movement and load specified by the user to the necessary muscle forces in a process known as inverse dynamics. Maximum synergism would be the case where all muscles capable of a positive contribution to balancing the external load work together, in such a way that the maximum relative load of any muscle in the system is as small as possible. It means that the body would maximize its endurance and precisely, this criterion might decide survival of the fittest in an environment where organisms are competing with each other for limited resources. So in AnyBody a minimum fatigue criterion way is employed because fatigue is likely to happen first in the muscle working on the maximum relative load, and it makes physiological sense that the body might work that way.

To implement dynamic analysis of human foot, a multiple-units foot and shank model was created in AnyBody Modeling System. As shown in Fig. 3, the coordinates of knee, ankle, and toe joints and muscle joint points were determined by measuring datum of the human lower limb (McMinn et al., 1982). The shank was also built in the model for attaching
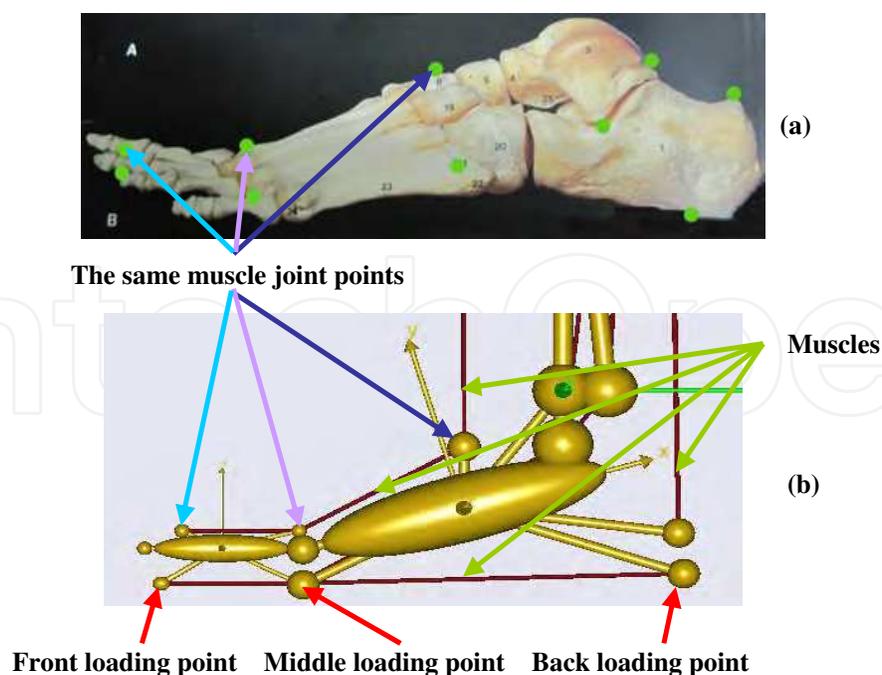

Fig. 3. Method for obtaining coordinates of multiple-units foot model in AnyBody System.
(a) Human anatomy sscannogram for obtaining geometric data of human foot.
(b) Musculoskeletal multiple-units foot model in AnyBody Modeling System.

muscles between foot and shank; furthermore the angular acceleration of the shank has influences on these muscles in the inverse dynamics method. To simulate the true situation of human walking, the GRF were implemented on three loading points locating on the bottom of foot model, and the loads applied on loading points contain both pressure and tangential forces as shown in Fig. 3 (b). The main working muscles involved in the model were ANT TIB, GAST, SOL, PL, Peroneus Brevis (PB), Posterior Tibialis (POST TIB), Extensor Digitorum Longus (EDL) and Flexor Digitorum Longus (FDL), that were built with the maximum strength of 5000 N. Totally seventeen muscle points were created to join muscles, four joints of hip, knee, ankle and toe were created to restrict the activity freedom degree, and three loading points were created to load the GRF and COP. The model worked as an integrated kinetics system after all the units were combined by joints and muscles. The foot model was established based on a rigid barefoot as the instrument shoes nearly have no elasticity, and to make the model suit for different individuals, the model size factor was indicated based on the length of human foot.

2.3 Inverse dynamics method for calculating muscle forces

The GRF, COP and angular motion data measured by the wearable sensor system were imported into the developed multiple-units foot and shank model in AnyBody Modeling System to calculate muscle forces of human foot through an inverse dynamic method. To simulate the true situation of human walking, one group of GRF and COP was transferred into two forces on different locations based on Eq. (1) and Eq. (2), in this way the measured two groups of GRF and COP data were transferred into three groups of force data applied on three loading points on the bottom plane of foot model. In single gait cycle, loads were transferred from posterior foot to the anterior foot continuously and the whole force process was simulated. This transformation can perfectly imitate the GRF variation and COP shifting because the multiple-units foot is regarded as a rigid body system, furthermore real human foot receives GRF mainly on similar three points too. The loads applied on loading points contain both pressure and tangential forces. The pressure force of GRF was expressed on the Z-direction, and the COP_z (x_0, y_0, z_0) on this dirction is defined by Eq. (1), and the tangential force of GRF was expressed on the Y-direction, and the COP_y (x_0, y_0, z_0) on this direction is defined by Eq. (2).

$$x_0 = 0 , \quad y_0 = \frac{\sum F_z \bullet y}{\sum F_z} , \quad z_0 = 0 \tag{1}$$

$$x_0 = 0 , \quad y_0 = \frac{\sum F_y \bullet y}{\sum F_y} , \quad z_0 = 0 \tag{2}$$

F_z — A force on Z-direction.
F_y — A force on Z-direction.
x — X-axis coordinate values of the force.
y — Y-axis coordinate values of the force.

As shown in Fig. 4, the angular motion data of shank, the posterior foot and the anterior foot were measured in general coordinate system as ϕ_{shank} , ϕ_{heel} , ϕ_{toe}. Because all the parts in the model were regarded as rigid bodies, the relative angular motion ($\Delta\phi_{shank}$, $\Delta\phi_{heel}$, $\Delta\phi_{toe}$)

could be calculated by quantitative subtraction. Furthermore in AnyBody Modeling System, the shank, the posterior foot and the anterior foot were driven by $\Delta\phi_{\text{shank}}$, $\Delta\phi_{heel}$, $\Delta\phi_{\text{toe}}$ respectively as shown in Eq. (3), (4), (5). After all the model units were combined by joints and driven by imported motion data, the AnyBody musculoskeletal model worked as an integrated kinetics system and quantitative muscle forces were calculated out through inverse dynamic process.

$$\Delta\phi_{\text{shank}} = \phi_{\text{shank}} \tag{3}$$

$$\Delta\phi_{\text{heel}} = \phi_{\text{heel}} - \phi_{\text{shank}} \tag{4}$$

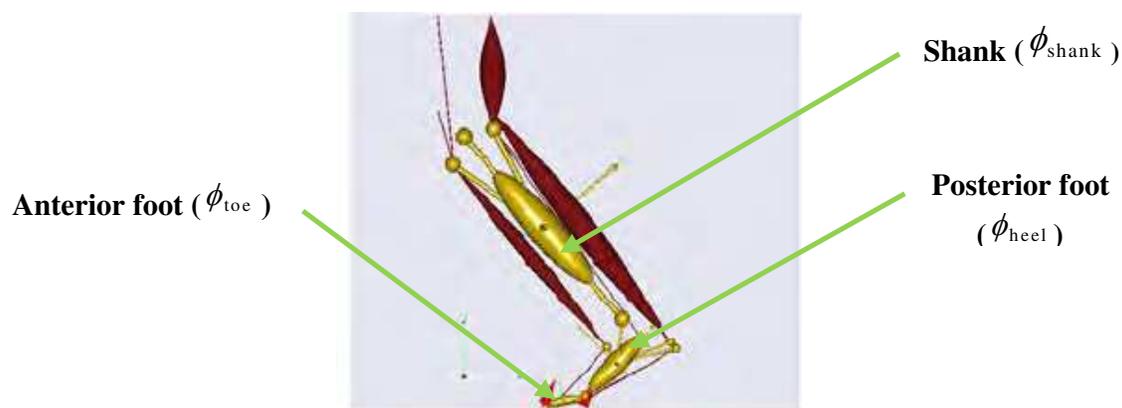$$\Delta\phi_{\text{toe}} = \phi_{\text{toe}} - \phi_{\text{heel}} \tag{5}$$

Fig. 4. Angular motion data of shank, the posterior foot and the anterior foot were measured in general coordinate system as ϕ_{shank}, ϕ_{heel}, ϕ_{toe}.

2.4 EMG method for validating muscle force results

The main muscles of human shank and foot were illustrated in Fig. 5, the motions of human limb are motivated by complex teamwork of these muscles, but only several of these muscles were involved in the EMG experiment, that because only the muscles visible in skin surface and offering primary motive power in walking process are possible and worthy for directly analysing. As shown in Fig. 6, the personal-EMG system (P-EMG-0403A01) includes hard type sensor system, filter box and data process system, and based on this system the EMG method was adopted to directly measure muscle activation level of ANT TIB, PL, GAST and SOL. The raw EMG results were filtered and rectified by the filter box and the data process system into integral EMG results that could represent the muscle activation levels. Both the raw EMG and the integral EMG were real time recorded and displayed by the data process system, furthermore the integral EMG results were introduced in the validation with muscle force results calculated by AnyBody Modeling System. Furthermore, one hundred percent standard voluntary contractions (100%SVC) were defined as standard isolation of muscle activity in the respective muscle tests for the normalization of EMG signal (Perry et al., 1986).
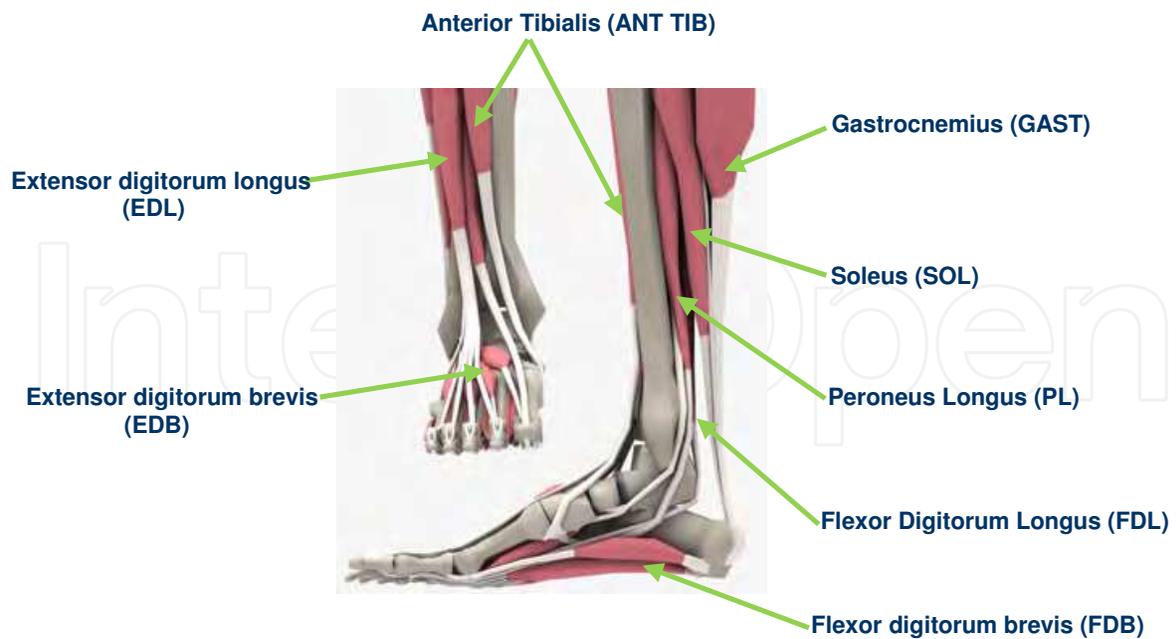
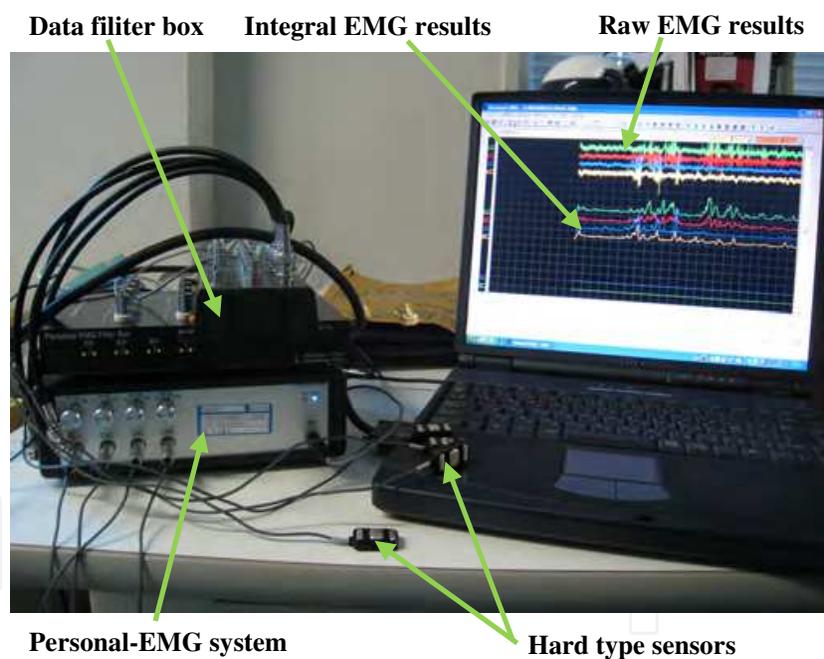Fig. 5. Illustration of main muscles of human shank and foot.

Fig. 6. Personal-EMG system adopted to directly measure muscle activation levels in human lower limb.

3. Experimental study

3.1 Experiment method

The first experiment step is acquisition motion and GRF information of human lower limbs in gait cycle, five adult volunteers (age: 28.5±5 years, weight: 75±6.5 kg.) who had no

musculoskeletal disease history were requireed to performed their normal gait in the experiment. The distance of the performance was four meters, duration time was ten seconds. As shown in Fig. 7, three-dimensional angular motion of thigh, shank, posterior part and anterior part of the multiple-units foot were measured with the unit of degree (°), and three-dimensional GRF and two-dimensional COP were measured by the instrument shoes with the unit of Newton (N) and millimeter (mm) respectively. Because all the limb parts were regarded as rigid segments, precise location of wearable sensors which attached on human limbs was unnecessary, so was the size adjustable of the instrumented shoes for the same reason. The sampling frequency of all sensors were regulated as 100 Hz, signal synchronization was realized by the signal process box which would send a pulse to wearable sensors attached on shank and thigh. To remove noise, low-pass filtering was performed on obtained signals with the cut-off frequency of 10 Hz.
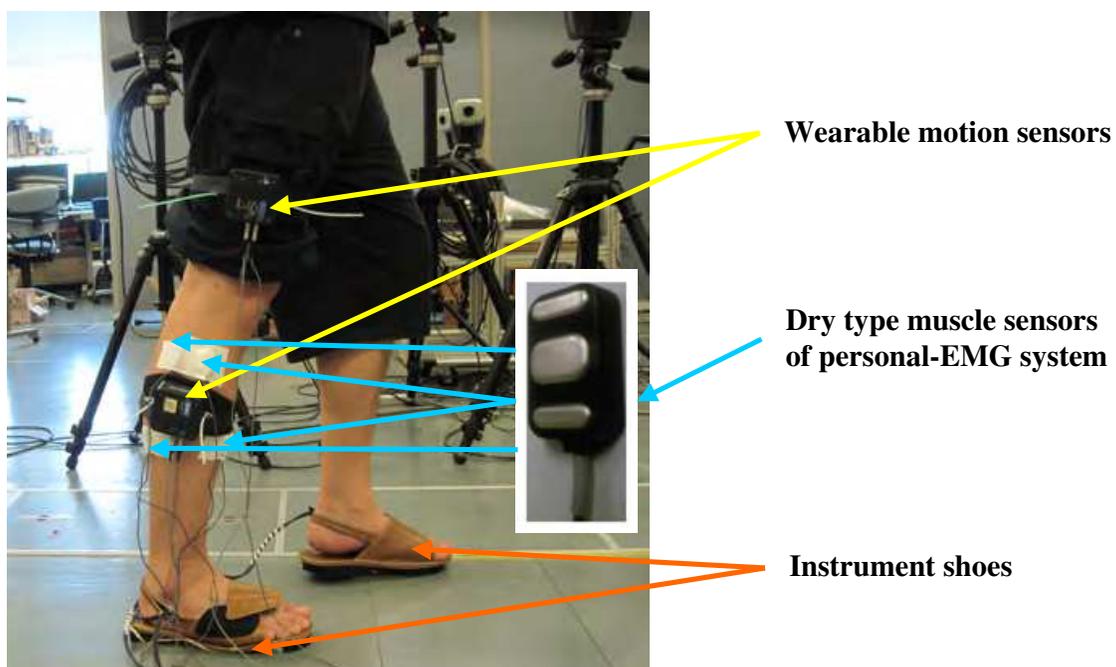

Fig. 7. Prototype of a volunteer wearing instrument shoes, wearable motion sensors and muscle sensors of personal-EMG in the experiment.

To directly measure muscle activation levels of ANT TIB, PL, GAST and SOL, the personal-EMG system and dry type muscle sensors were adopted in the experiment. The dry type muscle senors, which works on a myoelectricity difference principle, could sensing the muscle activation level while being pasted on the eneurosis zone of muscle surface. Furthermore the measured raw EMG results were real time filtered and rectified into integral EMG results by the personal-EMG application system, and both raw EMG results and integral EMG results were real time displayed for easy checking and adjusting. As shown in Fig. 8, we divided one gait cycle into four steps: contacting step, supporting step, leaving step and swing step. The contacting step starts from the heel contacting the ground and ends in the whole foot bottom plane stamping on the ground, the supporting step represents the process of gravity movement from posterior to anterior while one foot supporting the whole weight, the leaving step stands for the action from the heel leaving the ground to the whole foot leaving the ground, and the swing step means the foot swing forward process without contacting the floor.

Gait Cycle

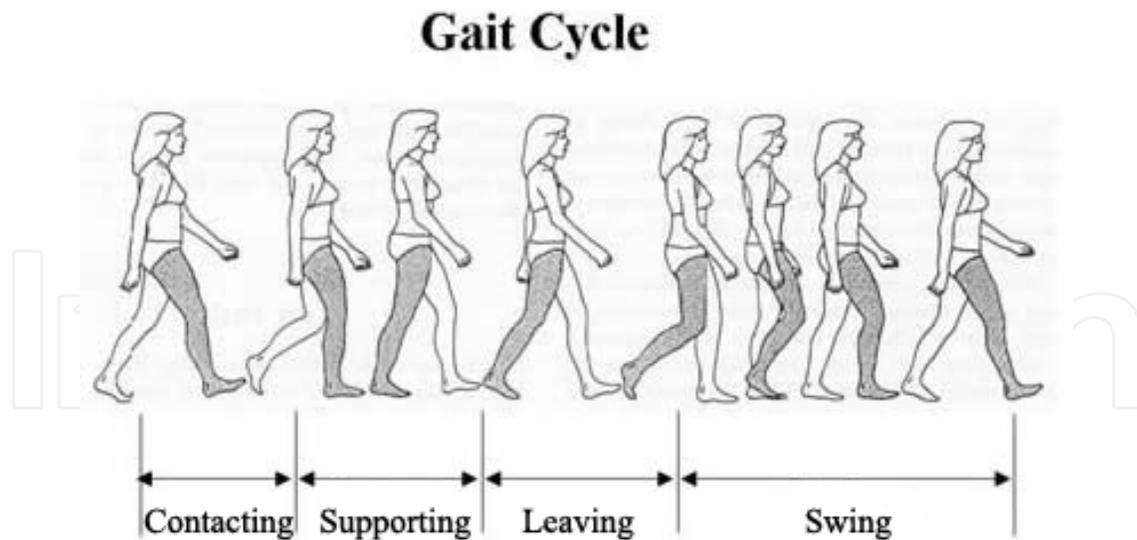

Contacting | Supporting | Leaving | Swing

Fig. 8. One gait cycle was divided into contacting step, supporting step, leaving step and swing step.

The second experiment step was calculating the muscle forces in AnyBody Modeling System. For preparation, the measured motion and GRF data of lower limb were transferred into the form that is adequate to the AnyBody system. As a rigid body system, the relative angular motion data ($\Delta\phi_{shank}$, $\Delta\phi_{heel}$, $\Delta\phi_{toe}$) were imported into the model to drive the movement of the shank and the multiple-units foot. The measured GRF data were transformed into three forces on three loading points to imitate GRF variation and COP shifting. As shown in Fig. 9, in Z-axis the entire pressure GRF data measured by sensors was drawn as F_z, and GRF

$$x_0 = 0$$
$$y_0 = \frac{\sum F_z \bullet y}{\sum F_z}$$
$$z_0 = 0$$

$$x_0 = 0$$
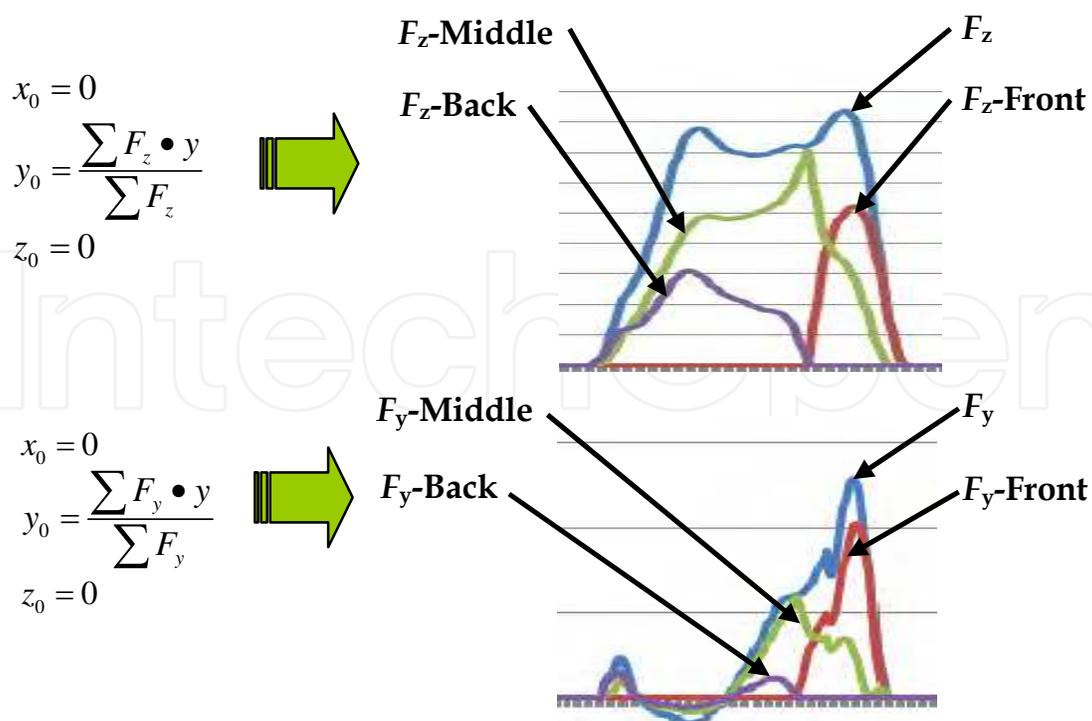$$y_0 = \frac{\sum F_y \bullet y}{\sum F_y}$$
$$z_0 = 0$$

Fig. 9. The GRF were transferred into three forces on three loading points to imitate GRF variation and COP shifting in gait cycle.

transformation data applied on three loading points were drawn as F_z-Front, F_z-Middle and F_z-Back. Futhermore in Y-axis, the entire tangential GRF data were transformed into F_y-Front, F_y-Middle and F_y-Back with the same algorithm. In order to make the foot model suitable for diferent individual, the size factors of foot length were dopted. As shown in Table 1, the size fators of five volunteers were summarized. Finally, the muscle forces and joint moment, while ankle joint muscles collaborating with each other in normal walking, were calculated through the inverse dynamics method in AnyBody Modeling System. This experiment has been approved by the ethics committee of the Department of Intelligent Mechanical System Engineering, Kochi University of Technology.

Subject	Size factor of AnyBody foot model (mm)
1	245
2	240
3	265
4	260
5	235

Table 1. List of size factors of the five subjects in the experiment.

3.2 Experimental results

The muscle force results through indirect AnyBody method and direct EMG method were summarized in this section. In AnyBody Modeling System the complex walking kinematic system requires high collaborations of all units, so the muscle forces of human foot were calculated based on a minimum fatigue criterion way. As shown in Fig. 10, the established AnyBody model, which can demonstrate the walking process and the muscle activation video, made the whole process easier to understand in a visualized way. The muscle activity level could be identified by the muscle bulge level, the value and direction of GRF could be discerned by the length and direction of the arrow in the AnyBody gait process video. The quantitative results of ANT TIB, POST TIB, GAST, SOL, EDL and FDL of five subjects in their normal gait were shown in Fig. 11, in Y-axis the quantitative muscle forces were expressed while in X-axis the time was represented. The muscles connecting shank and foot were classified in foot muscle analysis because these muscles would offer foot driven power, whereas the muscle from thigh to shank were excluded because those muscles could only drive the movement of shank in the gait dynamitic system.

To validate the muscle forces calculated by the musculoskeletal multiple-units foot model, contradistinctive analysis was implemented between AnyBody results and EMG results. Only ANTTIB, PL, SOL and GAST were involved in the analysis becasue these four muscles are powerful muscles for driving the movements of human limbs in normal gait, futhermore these muscles are easy to find in skin surface for surface EMG measurement. As shown in Fig. 12, the AnyBody results were drawn in blue while the EMG results were expressed in shadow, in comparison diagrams the X-axis represented percentage of gait cycle (%GC) while the left Y-axis indicated the muscle force with the unit of Newton, and the right Y-axis indicated the percentage of standard voluntary contraction (%SVC) of muscles. Futhermore for making the discussion clearly, the step division figure of gait cycle was expressed on the top of each diagram.
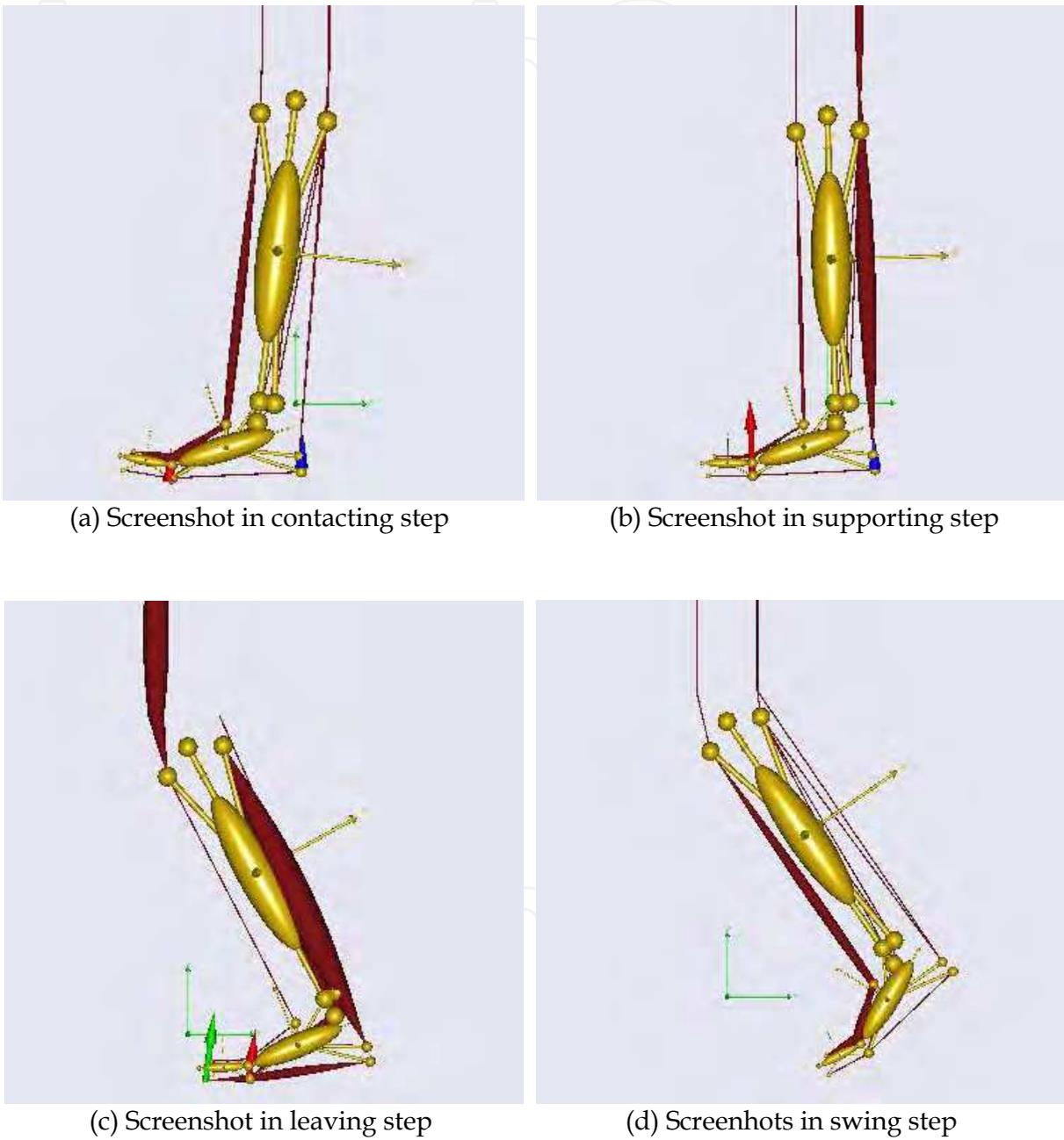
(a) Screenshot in contacting step   (b) Screenshot in supporting step

(c) Screenshot in leaving step   (d) Screenhots in swing step

Fig. 10. Screenshots of gait video in AnyBody Modeling System, four postures in one gait cycle were shown.

(a) Muscle forces diagrams of ANT TIB
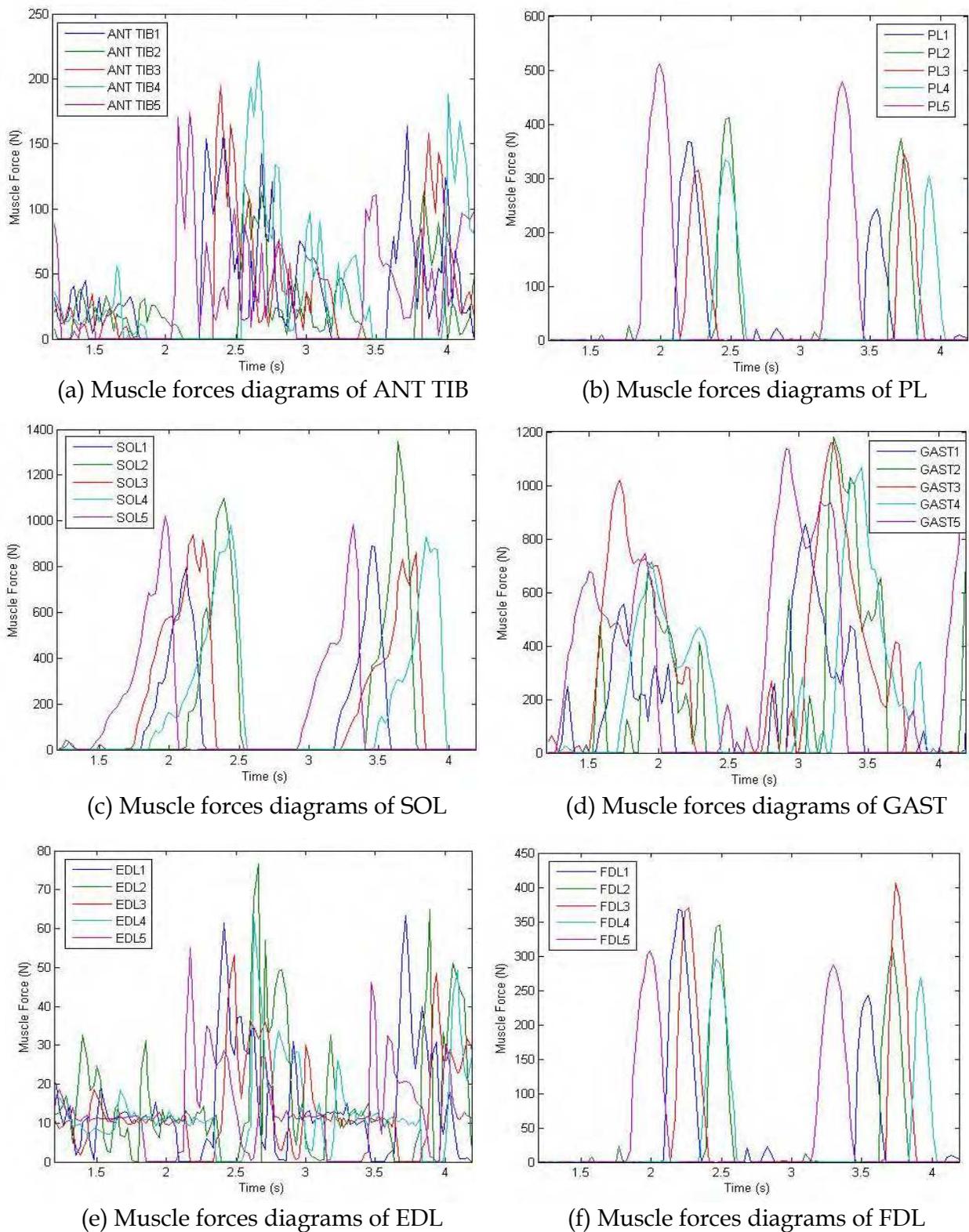
(b) Muscle forces diagrams of PL

(c) Muscle forces diagrams of SOL

(d) Muscle forces diagrams of GAST

(e) Muscle forces diagrams of EDL

(f) Muscle forces diagrams of FDL

Fig. 11. Muscle forces diagrams of ANT TIB, PL, SOL, GAST, FDL and EDL of five subjects obtained in AnyBody Modeling System.
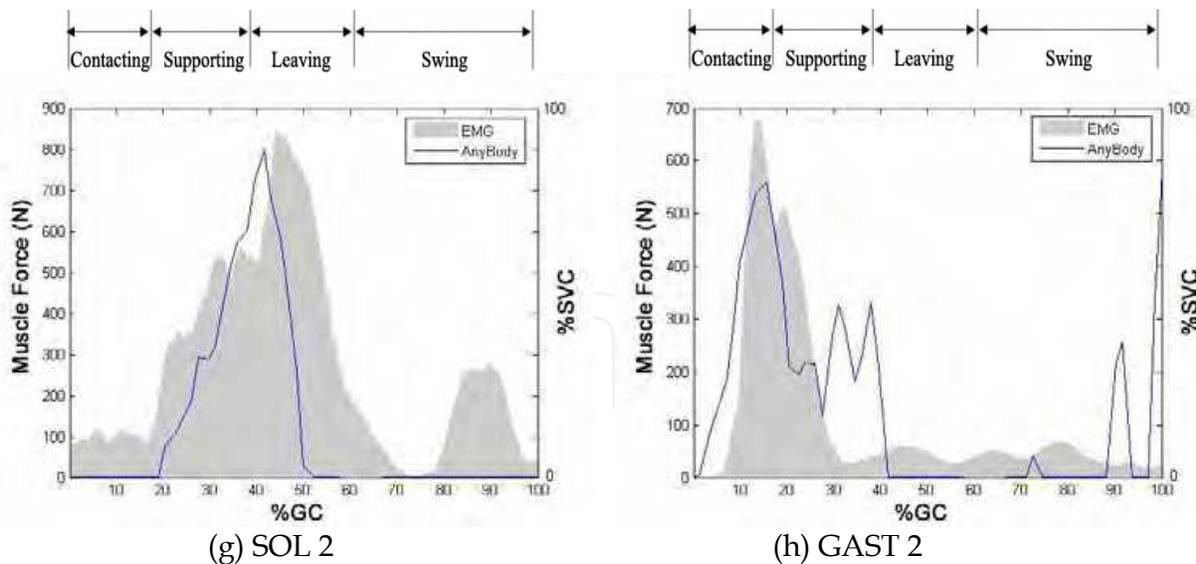
(a) ANT TIB 1

(b) PL 1

(c) SOL 1

(d) GAST 1

(e) ANT TIB 2

(f) PL 2

Fig. 12. (Continued)

(g) SOL 2   (h) GAST 2
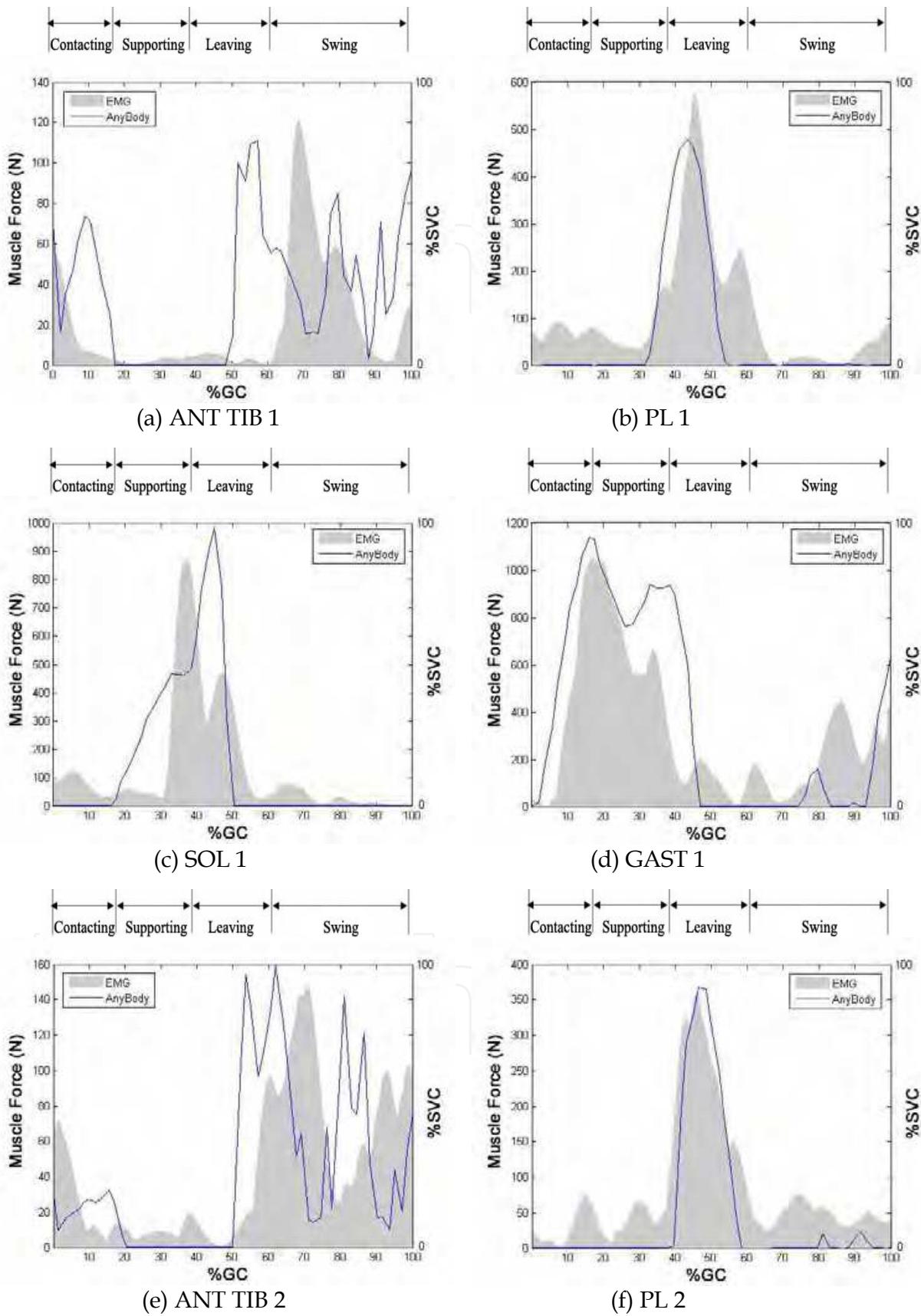
Fig. 12. Comparison diagrams of muscle activation of ANT TIB, PL, SOL and GAST in one gait cycle. The results of AnyBody Modeling System were drawn in blue while the results of EMG method were expressed in shadow.

4. Discussions

As muscle force results were estimated base on wearable sensor system and AnyBody Modeling system, EMG results had been used to do the comparison as in EMG method the muscle activity result is relatively large or small. As shown in Fig. 11, AnyBody results of five volunteers was exhibited, futhermore reality sense for walking process and similarty could be found among different indivaduals. And in Fig. 12, contradistinctive analysis was implemented between calculated results of AnyBody method and measured results of EMG method, obviously similar comparability could be found although the two results had many differences.

As shown in Fig. 11 (a), (e), the muscle force of ANT TIB and EDL were relatively larger in the first contacting step because the muscles in fronter of shank need to grow up to balance the moment of ankle joint while the heel contacting the floor. And in the last swing step ANT TIB and EDL were larger because they offered forces to drive the limb swing till next gait cycle started. Whereas the values of ANT TIB and EDL were keeping nearly zero in the other steps of gait cycle. As shown in Fig. 12 (a), (e), the value of ANT TIB was relatively high in both AnyBody results and EMG results in the beginning and ending period of one gait cycle, but in the middle steps the AnyBody results were keeping nearly zero while the EMG results were only keeping relatively smaller values.

As shown in Fig. 11 (c), (d), the tensile forces of SOL and GAST increased obviously in the supporting step when the GRF were transfered from posterior foot to anterior foot, moreover the peak value of GAST occurred in this step too. And in the leaving ground step, the SOL continued to increase for pushing human body moving forward and upward till the whole foot separated from the ground and start to swing, moreover the peak value of SOL occurred in this step. The peak Achilles tendon forces occured in the leaving step of our results that ranging from 1.07 kN to 1.66 kN, which were in good agreement with the reported peak Achilles tendon forces of 1430±500 N (Finni et al., 1998). The SOL and GAST were keeping small vaules in the contacting step because the center of gravity were behind

the ankle joint. As shown in Fig. 12 (c), (d), (g), (h), the AnyBody results and EMG results were not in high agreement but the variation tendencies were similar.

As shown in Fig. 11 (b), (f), FDL and PL had similar rising period and descending period, and peak values of PL and FDL both occurred in the leaving step, that because the two muscles are both located in the posterior of shank and play a similar role of pushing human body moving forward and upward in the normal human gait. As shown in Fig. 12 (b), (f), in contacting, supporting and swing steps the EMG results of PL showed relatively higher values while the AnyBody results were keeping nearly zero, but in the whole gait cycle obvious comparability could be found between the two results.

5. Conclusions

Muscle forces of human foot were estimated through the inverse dynamics method in AnyBody Modeling System based on wearable sensors and EMG method. The angular motions, GRF and COP of human limb segments could be measured by the developed wearable motion and force sensors with acceleration sensitive units and gyroscopes in various situations. AnyBody Modeling System that professional concerns on musculoskeletal kinematics and kinetics analysis could imported the sensor measured data into a bionic dynamic model. And through invise dynamic method the quantitative muscle forces of human foot could be estimated in AnyBody system. The EMG method was adopted to directly measure muscle activition levels for validation, and the muscle activation results of EMG method have certain comparability with the muscle force results of AnyBody model, futhermore the tendency curves of these muscle forces have reality sense for walking process analysis. This reserch method for human foot appears to be a practical means to determine muscle forces in musculoskeletal analysis and in on-the-spot medical applications. In the future, the muscle force can be made into vector quantity instead of scalar quantity by 3D technology in AnyBody Modeling System as the angular motion and GRF sensors own the capability of 3D measurement.

6. Acknowledgement

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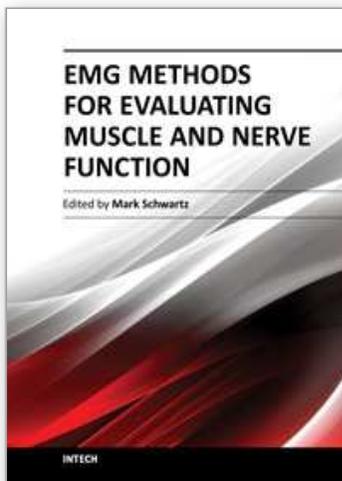
EMG Methods for Evaluating Muscle and Nerve Function
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