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ORIGINAL ARTICLE

Reconstruction of walking motion without flight phase by using computer simulation on the world elite 20 km female race walkers during official race

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ABSTRACT

BACKGROUND: The aim of this study is to obtain basic findings of the walking motion without flight phase. From three- dimensional coordinate data of athlete in the real official competition, virtual three-dimensional coordinate will be calculated, and the joint kinetics will be obtained by using inversed dynamics.
METHODS: Two kinds of computer simulation were conducted on the three-dimensional whole body coordinates data of twenty-two world elite female walkers collected in the elite official race (World championships in Athletics held in 2005 and 2007) for Women's 20 kmW. In most of elite race walkers, their recovery phase time was larger than the support phase time. One simulation directed to diminish the flight phase by setting the recovery phase same as the support phase of contralateral leg. Another simulation directed to diminish the flight phase by setting the start of the support phase as the start of the recovery phase of contralateral leg.
RESULTS: In the simulation, which directed to reduce the recovery phase time, the walking speed was same as the real data. However, the larger joint torque at hip and knee exerted rather than the real data. In the simulation, which directed to expand the support phase time, the walking speed was same as the real data.
CONCLUSIONS: The large joint torques at ankle and knee in the final phase of the support phase exerted rather than the real data and helps large forward swing velocity in the early recovery phase with lower knee extensors' and hip flexors' torque rather than real data.

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Key words: Computer simulation; Joints; Torque.

Article 230 of competition and technical rules of World Athletics (WA) defines the race walking such that walker should contact the ground and that the advancing leg to be straightened from the moment of the first contact with the ground until the vertical position of the support leg.[1] There should be a minimum of six judges and a maximum of nine judges on the circuit course up to two kilometers. Their judgement shall be based on observations by human eyes, not detections with electric nor optical devices. Athletes who are advocated as to violate definition of race walking by three different judges; they must be disqualified or required to entre Penalty Zone during an applicable period along with race distance, two minutes for 20kmW and five minutes for 50 kmW. In the education and certification system of race-walking judges, WA divides the international race-walking judges into three different categories, Level I as Domestic, Level II as Area (Continent), and Level III as World.[2] In the world athletic series of WA, as Olympics, the world championships, and the world race walking team championships, only Level III judge shall be appointed. In the area competition, Level II and III shall be appointed. Level I can be appointed in other kind of international competition. After 2005, WA limited the world championships and Olympics to the athletes who marked performances below the entry standard at approved competitions with at least three Level II or III judges. This system has aimed to avoid disqualification in the high-level international competitions, the world championships, and Olympics. However, after 2005, disqualification still occurs in the world championships and the Olympics, nevertheless all participants have not been disqualified in the domestic qualification where Level II and III judges has judged. Because it should be difficult to maintain the technique and fitness of athlete, the violation from the definition may occur in the high-level international competitions. That suggest that any level of competitions need to be controlled by race walking judges. In recent international competitions, especially men's 20km race walking, where athletes compete in high walking speed, most of three red cards are issued in the claim of loss of contact by three different judges. All three red cards issued in the claim of bent knee are very few. Then, most of problems in judgment on major competition are likely the judgments of loss of contact. Article 230 of competition rules specify that race walking should be judged by human eye of race-walking judges, but even highly skilled race-walking judges may sometimes not recognize a brief "loss of contact" time during race. The critical walking speed at which walkers lose contact with the ground in experiments was reported.[3-10] In these facts, the definition of race walking in competition rule shall conflict with the practical judges. As previous research, flight phase has occurred in most of walking races in which elite athlete participated.[3-11] It is difficult to find out elite race walker who has walked without flight time. As the photo finish system has been applied to the athletics of Olympics in London, 1948, the electric devices have been demanded to be applied to race walking for long time. In France of 1920's and in Soviet Union of 1950's, sensors for loss of contact were developed only for training.[12] IAAF has announced the start of discussion to apply Sensor shoes with built-in sensor to the judgment of loss of contact after Sydney Olympic games.[8] However, these devices have not been applied to the official competition because of problems of practical uses. However, IAAF has held a technical seminar for athletes, coaches, judges, and other officials during the world race walking team championships in 2018, where the prototype of electric judging system for the future was presented.[13] Although some aspects of technical rules should be adjusted to apply this system to the practical competitions, the innovators of this system which can detect the length of loss of contact time during entire races mentioned it to expand the style of competition management of this event including judgement. Because of these proposals on race walking rules and its technical management, it is appropriate to apply computer simulation methods to establish virtually walking motion which is consistent to the definition of competition rule for the training and coaching for athletes. There are various purposes and merits of application of computer simulation to the research of human motion.[14, 15] To develop new-technique and enhance performance is one of effective merit of computer simulation to contribute to the research of high-performance sports.[16-18] The strength of computer simulation modelling for sports science support is that it can provide general research results for the understanding of elite performance.[15] There is another merit of this methods, that the influence of the fluctuation of motion and output of muscle contraction can be tested with real attempt of athletes.[19] In the computer simulation of human body, many forward dynamics, in which the human motion in the fluctuated human muscle output are calculated, are conducted. However, the inverted dynamics are also conducted.[20] In the simulation to the optimal solution and motion with forward dynamics, there is a problem how to evaluate the probability of optimal solution and the objective function.[17, 20] Then, in race walking, it is appropriate to discuss with the solution of inverted dynamics by using the model without flight phase. Hoga-Miura *et al.*[2] conducted reconstruction of simulated motion of race walking without flight phase, based on the data obtained in the real official competition. They also calculated kinetic data with inversed dynamics from the simulated data. In the result of the simulation which aimed to reduce the recovery phase time, the walking speed was reduced because of the short step length. However, the larger joint torque at hip and knee exerted rather than the real data. In the result of the simulation which aimed to expand the support phase time, the walking speed was same as the real data. Although the large joint torque at hip and knee exerted rather than the real data, the hip abductors' torque which was large in the walkers without flight phase in the official race and experiment was reduced in the simulated data. The simulation in the frontal plane should be conducted in the future. Although the competition rule in race walking for women's events are all the same as for men's event, the number of red cards and disqualification on women's event were smaller than that on men's event. As there are differences between men and women in the mechanical property for higher walking speed in race walking, the mechanics to occur loss-of contact phase might be difference between men and women. The aim of this study was to obtain simulated motion of race walking without flight phase in women's event. From three- dimensional coordinate data of athlete in the real official competition, virtual three-dimensional coordinate will be calculated, and the joint kinetics is obtained by using inversed dynamics.

Materials and methods

Data collection and processing

Data were collected (under a grant from the Scientific Research Committee of Japan Association of Athletics Federations) on three official men's 20-km race walking competitions held on 2-km courses: World Championships in Athletics held in 2005 (Helsinki) and 2007 (Osaka). Walkers passing along the straight and flat sections of the courses were videotaped at 60 Hz with two video-tape recording (VTR) cameras positioned at right sides of the course, as shown in Figure 1. A calibration matrix (3.0×4.5×2.0 m) for the three-dimensional transformation method was set on the walking course with markers on a rigid steel pole and videotaped before and after the races. Images from the two cameras were synchronized with a right foot contact frame. Appointed international race-walking judges controlled all competitors in the present study under regulations of WA. Twenty-two walkers (Table I; fifteen from Helsinki; seven from Osaka) were selected as subjects, who were visible on both two cameras, finished below 1:37:00 and had never disqualified on the international competitions before and after two years of the analyzed races. About athletes who walked both competitions, the better performance of each athlete was selected for the analysis. In the images from the two cameras on the right, 25-endopoints of each subject were digitized throughout one walking cycle at 60 Hz from the VTR images using digitization software (Frame-DIAS-II, DKH Inc., Japan). The coordinates of the digitized points were converted into real three-dimensional coordinates using direct linear transformation (DLT) method and then smoothed with a Butterworth low-pass digital filter. Optimal cut-off frequencies were determined by the residual error method proposed by Wells and Winter.[21] These ranged from 1.2 to 8.4

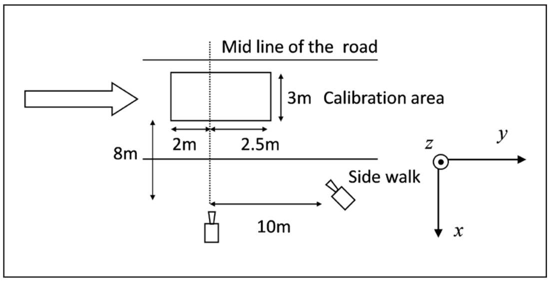


Figure 1.—VTR cameras set-up.

TABLE I.—*Characteristics of the subjects (N.=22).*

Variables	Mean±SD
Age (yrs)	27.4±4.30
Height (m)	1.66±0.04
Body mass (kg)	52.7±5.1
Race record	$1_h31_m27_s\pm2_m53_s$
Personal records before race	$1_h28_m57_s\pm2_m03_s$
Performance ratio (%)	97.3±2.7

Hz for the medial-lateral coordinates, from 1.8 to 8.4 Hz for the anterior-posterior coordinates and from 3.0 to 8.4 Hz for the vertical coordinates. A three-dimensional 19-segment model comprised of the hand, forearm, upper arm, foot, shank, thigh, leg (vector from hip to ankle), head, upper torso (longitudinal axis), lower torso (longitudinal axis), shoulder girdle (vector from right to left shoulder) and pelvis (vector from right to left hip) was used to calculate the linear and angular kinematics of joints and the centers of mass of each body segment of the subjects as Hoga *et al.*[22] In addition, the leg angle of both right and left leg was obtained as the angle from the vertical downward vector to the vector from hip joint to the ankle joint. The location of the centers of gravity of the whole body (CG) and the masses of the subjects' body segments were estimated from the body segment parameters of Ae *et al.*[23] A walking cycle was defined as the period from the instant of right toe-off to the instant of the next toe-off. Step frequency was determined as the reciprocal of one half of the time elapsed for a cycle. Step length was determined as one half of the distance that CG progressed during one walking cycle. The walking speed for one step was calculated as the product of the step length and the step frequency. The competition rules of WA specify that race walking should be judged by eye by the race walking judges, but during races even highly skilled race-walking judges may sometimes not recognize a brief "loss of contact" time.[3-11] In the present study, one walking cycle was divided into the support phase and the flight phase. The support phase was defined as at least one foot keeping contact with the ground and the flight phase was defined as both feet losing contact with the ground. One half of the time for the support phase during one walking cycle was defined as the support time, and one half of the flight phase time was defined as the flight time. The horizontal distance during one support phase was defined as the support length, and the distance during the flight phase was defined as the flight length.

Simulation

In the present study, computer simulations have been conducted with the methods of Hoga-Miura *et al.*[2] as follows. The measured data (real data) was simulated into two kinds of models. In the first model (simulation A in Figure 2), the time of recovery phase was shortened to the time of support phase of antagonist leg. The second one (simulation B in Figure 3) was that the support length was elongated to the time of the recovery phase of antagonist leg. The process of Simu (simulation) A was conducted as follows:

- process 1 – from the three-dimensional coordinate raw data of the twenty-five endpoints for each subject, the relative coordinates of recovery leg (toe, ball, heel, ankle, and knee) to the recovery hip during the first step recovery phase from one frame before the first step toe-off ($T_{T\text{-}off1}$) to the heel on ($T_{H\text{-}on2}$) was calculated (Figure 2A);
- process 2 – the time of the recovery phase for real data was converted into the time of recovery phase for Simu A: from the Heel-on of the antagonist leg ($T_{H\text{-}on1}$) to the one frame before to the Toe-off of the antagonist leg ($T_{T\text{-}off2}$). The simulated relative coordinates of recovery leg were calculated by using interpolation of spline function. In addition, the simulated relative coordinate data was added to the recovery hip joint coordinate data. For the second step, same calculation as the first step was conducted (Figure 2B);
- process 3 – the difference of the anterior-posterior coordinates of the centre of gravity (D_{flight}) from the toe-off of the second step as the start of the second step ($T_{H\text{-}on2}$) to the toe-off of the first step as the end of the first step ($T_{T\text{-}off2}$) was calculated. This D_{flight} was detracted from the anterior posterior coordinate from the entire phase of second step to obtain the whole-body coordinate of the second step. After the present simulation the time point of $T_{H\text{-}on2}$ and $T_{T\text{-}off2}$ become the same time point, the whole-body coordinate of the end of the first step and the second step was averaged as the whole body. The initial time point of the first step was defined as $T_{initial1}$ and the second step as $T_{initial2}$ and the end of the second step as T_{final2} (Figure 2C);
- process 4 – the simulated data was smoothed with a Butterworth low-pass digital filter same as raw data (Figure 2C);

The process of Simu B was conducted as follows:

- process 1 – the three-dimensional coordi-

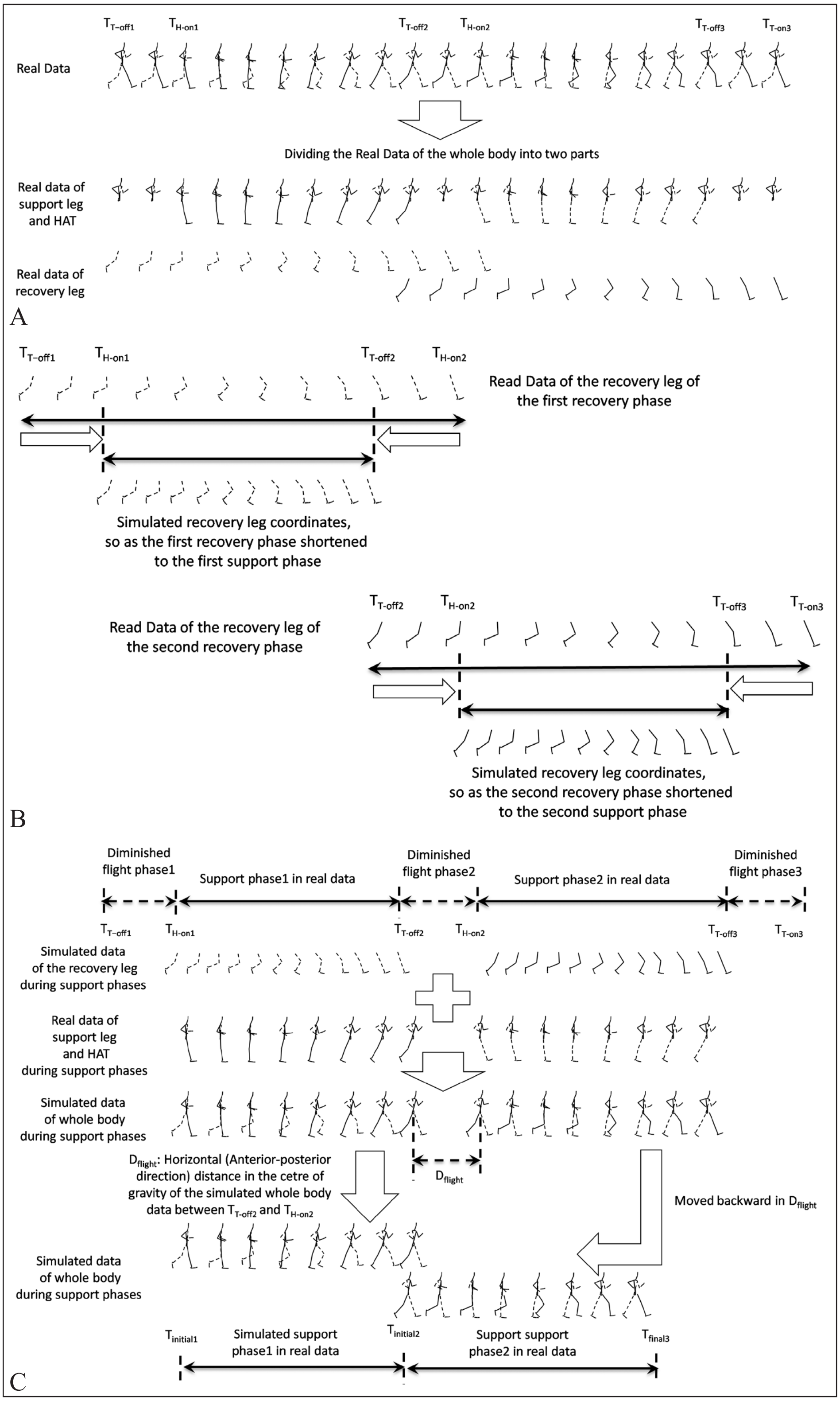


Figure 2.—A) Process of calculation for Simu A: dividing the Real data of the whole body into two parts, support leg and HAT, recovery leg; B) process of calculation for Simu A: shortening of the time of recovery phases into the support phase of the antagonist leg; and C) process of calculation for Simu A: reconstruction of the simulated whole body data.

nate of the end point of the first step support leg (toe, ball, heel, ankle, knee, hip) was converted with the rotation matrix about the heel coordinate at the heel on time of the first step ($T_{H\text{-}on1}$) in the sagittal plane in the clockwise. The range of rotation was until the anterior-posterior coordinate of the support hip was to match the one frame before recovery leg toe-off ($T_{T\text{-}off1}$). The rotated coordinate of the support leg defined as support leg coordinate at the $T_{T\text{-}off1}$. As first step, the support leg of second step and the third step was calculated (Figure 3A);

- process 2 – the coordinates during the expanded support phase between $T_{T\text{-}off1}$ and $T_{H\text{-}on1}$, between $T_{T\text{-}off2}$ and $T_{H\text{-}on2}$ were calculated with the interpolation by using spline function (Figure 3A);
- process 3 – the relative coordinates of recovery foot and shank (Toe, ball, heel, ankle, and knee) to the recovery hip during the first step recovery phase from one frame before the first step toe-off ($T_{T\text{-}off1}$) to the heel on ($T_{H\text{-}on2}$) was calculated. The simulated relative coordinates of recovery foot and shank was calculated to the recovery phase time converted to the toe-off1 of the antagonist leg ($T_{T\text{-}off1}$) to the one frame before to the Toe-off of the antagonist leg ($T_{T\text{-}off2}$) by using interpolation of spline function. In addition, same calculation was conducted from the one frame before the toe-off of the second step ($T_{T\text{-}off2}$) to the heel on of the third step ($T_{H\text{-}on3}$). The expansion rate of simulated segment angles was calculated as follows: the ratio of the angle from one frame before toe-off to the simulated heel on to the angle from one frame before Toe-off to the heel on of the real data. The range of motion was expanded with that ratio (Figure 3B).

Assumption of ground reaction force

In the present study, the measurements of the ground reaction forces were not conducted because the force plat form could not be set on the official courses. However, the estimation of the ground reaction forces during official races of athletics were attempted by using the acceleration of the center of gravity and the angular momentum of the whole body.[2, 24-27] Hoga[25] has reported that the estimated ground reaction force by using the acceleration of the center of gravity and the angular momentum of the whole body was almost similar to the measured ground reaction force. Also, Hoga *et al.*[27] reported that the lateral component of the estimated data was similar to the measured. In the present study, the ground reaction forces were estimated on the method of Hoga *et al.*[27] to calculate the joint torque and torque power of the whole body. Anterior-posterior component of ground reaction force which was estimated as the product of the acceleration of the center of gravity and the body mass was largely different from the measured ground reaction force in running[26] and race walking.[25] However, vertical component of the estimated ground reaction force was not different.[25, 26] In the present study, only the vertical (Z) component of the ground reaction force (GRF_Z) was calculated from the acceleration of the center of gravity as in Equation 1:

$$GRF_Z = Ma_Z + Mg$$

where M indicate the mass of the whole body, a_z indicate the vertical component of the acceleration of the center of gravity, *g* indicates the gravitational acceleration. The moment about the center of gravity which was applied by the external force was equivalent to the differentiation of the angular momentum of the whole body about the center of gravity. Enomoto *et al.*[24] estimated the anterior-posterior component (Y) of the ground reaction force ($mGRF_Y$) by using the angular momentum of the whole body as Equations 2-4.

$$H_{CG} = \sum_{i=l}^{s}(I_i\omega_i + m_i r_{iy} rv_{iz} - m_i r_{iy} rv_{iz}$$

$$\dot{H}_{CG} = d_y GRF_Z - d_z GRF_y$$

$$mGRF_Y = \frac{d_y GRF_Z - \dot{H}_{CG}}{d_z}$$

where $\dot{H}_{CG}$ indicate the angular momentum of the whole body about the center of gravity, s is the number of segment, I_i is the momentum of inertia of the segment i about the center of mass, ω_i is the angular velocity of the segment i about the center of mass, m_i is the mass of the segment i, r_i is the position vector of the center of mass of the segment i to the center of gravity of the whole body, rv_i is the velocity vector of the center of mass of the segment i to the center of gravity of the whole body, $mGRF_Y$ is the anterior-posterior (Y) component of the ground reaction

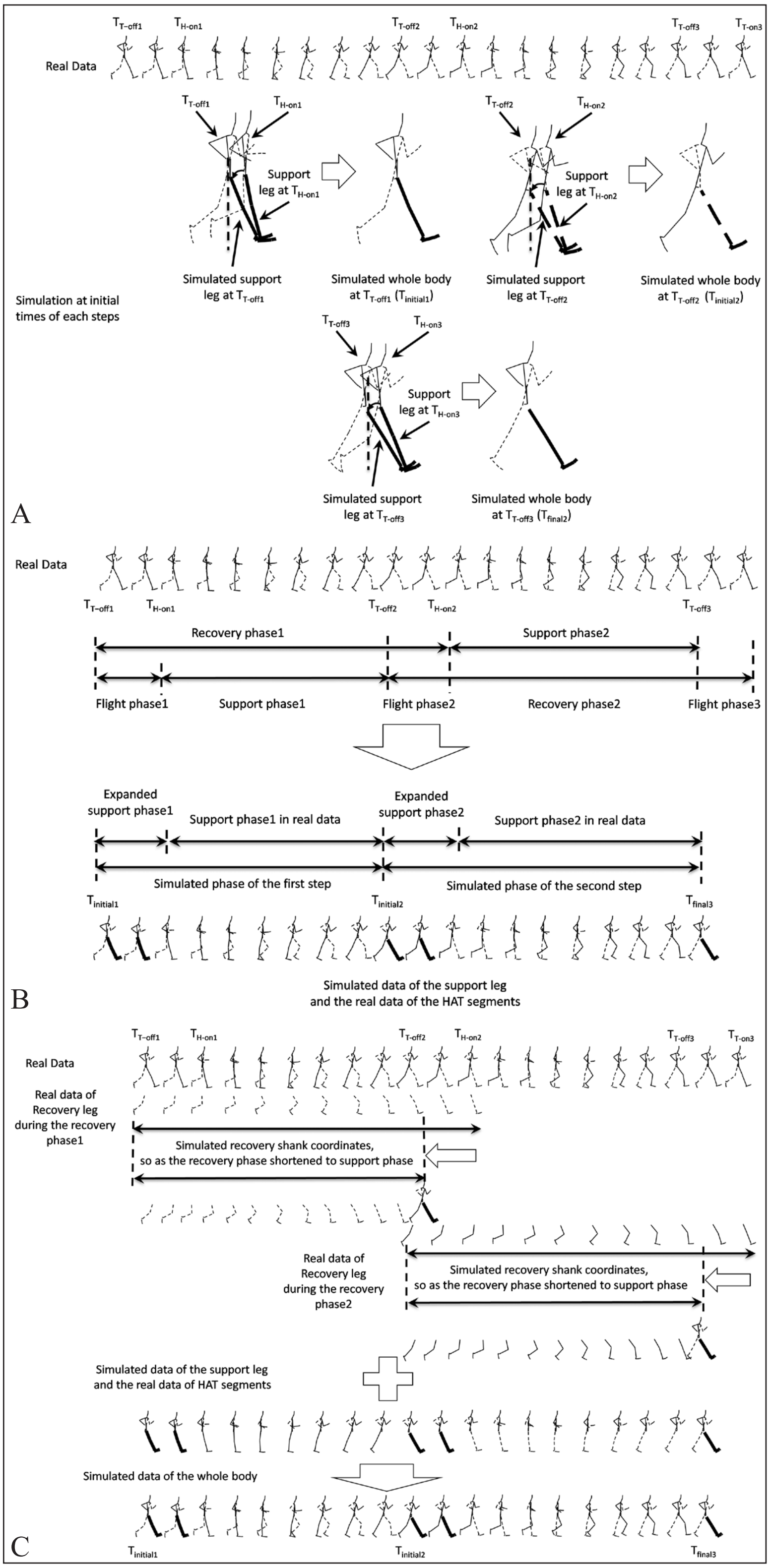


Figure 3.—A) Process of calculation for Simu B: simulated coordinates of the Heel-on; B) process of calculation for Simu B: simulated coordinates of the expanded support phase; and C) process of calculation for Simu B: reconstruction of the whole body simulated coordinates of the expanded support phase.

force which was calculated from the moment about the center of gravity of the whole body, d is the position vector of the center of pressure on the support foot to the center of gravity of the whole body. In the present study, the position of the center of pressure on the support foot was estimated by using the methods of Hoga[25] and Hoga et al.[27] The estimation was based on the assumption that the center of pressure was moved from the support heel to toe in the same velocity during the support phase.

Joint torque

Joint force and torque applied at the joints of the whole body were estimated as the solution of the equations of motion in the assumption that the human body was the link of the rigid body.[28]

Statistical analysis

Pearson's coefficient of correlations was calculated to test the relationships of the walking speed and the step parameters both in the measured data and the simulated data. The level of significance was set at 5%. The time series data were normalized by the time of the support phases, which was defined from RH-on to RT-off as 100% for the right foot support phase, and from LH-on to LT-off as 100% for the left foot support phase. Also, the data of the right foot support phase and the left support phase (H-on to T-off) were averaged to compare magnitudes and patterns between conditions. Paired *t*-test was conducted to compare the measured data and the simulated data. The level of significance was set at 5%.

Results

Real data

Table II shows walking speed and performance descriptors (step frequency, support time, flight time, step length, support length, and flight length) for all subjects at the analyzed points of each subject. Walking speed, support time, step length and flight length were significantly related to race speed. To the walking speed, support time, flight time, step length and flight length have significant relationships. However, walking speed was not significantly related to support length.

TABLE II.—*Step analysis for the Real data (N.=22).*

Variables	Mean±SD	Coefficient of relationships to race speed (*r*)	Coefficient of relationships to walking speed (*r*)
Walking speed (m/sec)	3.72±0.16	0.76*	
Step frequency (Hz)	3.34±0.12	0.30	0.36
Step time (sec)	0.30±0.01	-0.29	-0.36
Support time (sec)	0.27±0.01	-0.44*	-0.54*
Flight time (sec)	0.04±0.01	0.42	0.51*
Step length (m)	1.08±0.05	0.51*	0.69*
Support length (m)	0.92±0.05	0.12	0.21
Flight length (m)	0.17±0.04	0.52*	0.64*

*P<0.05.

Simulated data

Table III shows walking speed and performance descriptors (step frequency, support time and support length) of simulated data (Simu A and B) for all subjects. In both Simu A and Simu B, there was not significant difference to real data in walking speed. Because the process of Simu A was to shorten the recovery time into the support time, the step time and step length of Simu A was significantly smaller than that of real data. Figure 4 shows the mean joint torque at ankle in sagittal plane of real data and simulated data (Simu A and B) during the normalized recovery and support phase for all subjects (N.=22). The positive value indicates the dorsi-flexors' torque and the negative value indicates the plantar-flexors' torque. The ankle joint torque for all data were very small during the recovery phase. Although the plantar flexors' torque for Simu A was significantly smaller than that for real data in the second half of the support phase, that for Simu B was significantly larger than that of real data. Figure 5 shows the mean joint torque at knee in sagit-

TABLE III.—*Step analysis for simulated data (N.=22).*

Variables	Simu A, mean±SD	*t*-value of Simu A *vs.* Real data (*t*)	Simu B, mean±SD	*t*-value of Simu B *vs.* Real data (*t*)
Walking speed (m/sec)	3.71±0.19	0.85	3.71±0.17	0.59
Step frequency (Hz)	3.89±0.22	18.43*	3.34±0.12	-
Step time (sec)	0.26±0.01	23.4*	0.31±0.01	-
Step length (m)	0.92±0.05	20.24*	1.08±0.04	-

*P<0.05.

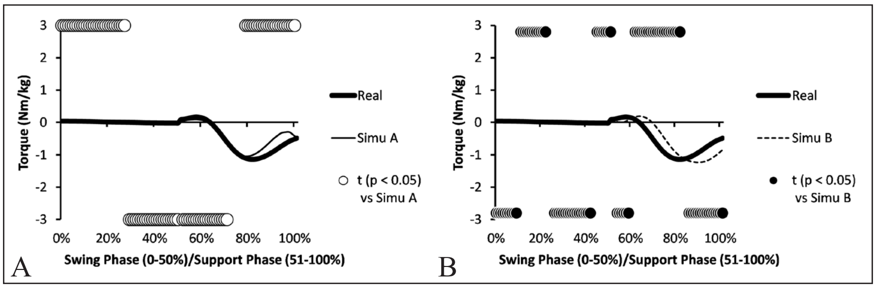


Figure 4.—Joint torque at ankle in sagittal plane of real data and simulated data (Simu A and B) during the normalized recovery and support phase for all subjects (N.=22).
Positive value (+): dorsiflexors' torque; negative value (-): plantar-flexors' torque.

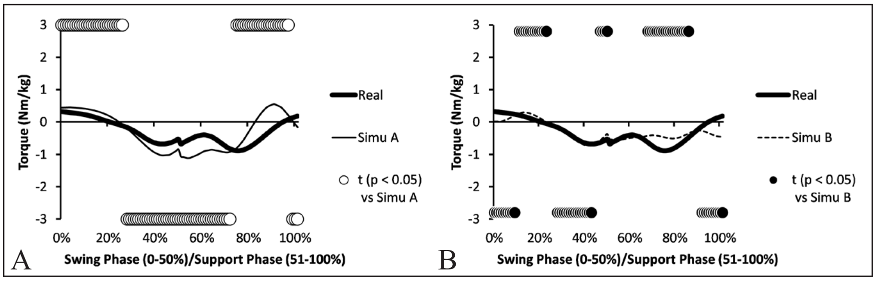


Figure 5.—Joint torque at knee in sagittal plane of real data and simulated data (Simu A and B) during the normalized recovery and support phase for all subjects (N.=22).
Positive value (+): extensors' torque; negative value (-): flexors' torque

tal plane. The positive value indicates the extensors' torque and the negative value indicates the flexors' torque. The knee extensors' torque was exerted in the first half of recovery phase and changed to knee flexors' torque in the middle of the recovery phase. During the recovery phase, both extensors' and flexors' torque of Simu A data was obviously larger than those of real data. However, the knee extensors' torque of Simu B in the initial part of the recovery phase were significantly smaller than that of real data. The flexors' torque of Simu B in the second half of recovery phase was significantly larger than that of real data as that of Simu A. During the support phase, although flexors' torque changed to extensors' in the end of the support phase of real data, that change occurred in the middle of the support phase of Simu A and the flexors' torque of Simu A in the first half of the support phase was significantly larger than that of real data. Although the flexors' torque in the middle of the support phase of Simu B was significantly smaller that of real data, that of Simu B was exerted continuously during the entire support phase. Figure 6 shows the mean joint torque at hip in sagittal plane. The positive value indicates the flexors' torque, and the negative value indicates the extensors' torque. The hip flexors' torque was exerted in the first half of recovery phase and changed to extensors' torque in the middle of the recovery phase. That torque changed to flexors' in the second half of the support phase. Both flexors' and extensors' torque of Simu A data were larger than those of real data in the recovery and support phase. However, in the recovery phase of Simu B, the flexors' torque in the initial part was significantly smaller than that of real data and the extensors' torque in the early second phase was significantly larger

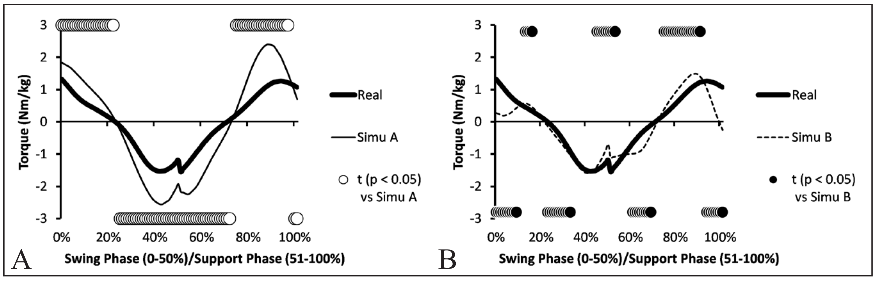


Figure 6.—Joint torque at hip in sagittal plane of real data and simulated data (Simu A and B) during the normalized recovery and support phase for all subjects (N.=22).
Positive value (+): flexors' torque; negative value (-): extensors' torque

Figure 7.—Joint torque at knee in frontal plane of real data and simulated data (Simu A and B) during the normalized recovery and support phase for all subjects (N.=22)
Positive value (+): varus torque; negative value (-): valgus torque

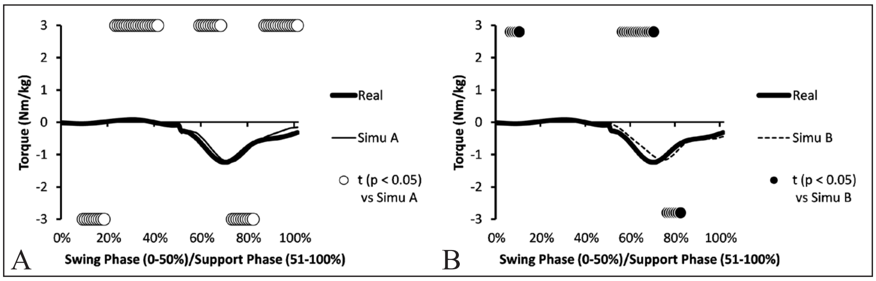
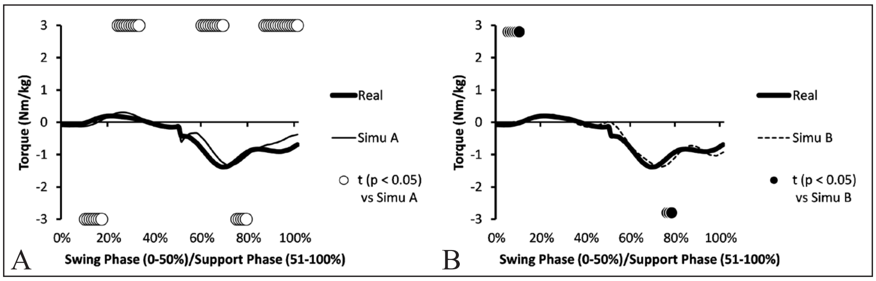


Figure 8.—Joint torque at hip in frontal plane of real data and simulated data (Simu A and B) during the normalized recovery and support phase for all subjects (N.=22)
Positive value (+): adductors' torque; negative value (-): abductors' torque.



than that of real data. The hip extensors' torque in the first half of the support phase and the hip flexors' torque in the second half of both Simu A and Simu B were significantly larger than those of real data. Figure 7 shows the mean joint torque at knee in frontal plane. The positive value indicates the varus torque, and the negative value indicates the valgus torque. Although those torques for all data were very small during the recovery phase, the peak of valgus torque of both Simu A and B were almost the same as that of real data in the middle of support phase. Figure 8 shows the mean joint torque at hip in frontal plane. The positive value indicates the adductors' torque, and the negative value indicates the abductors' torque. Although those torques for all data were very small during the recovery phase, the peak of the abductors' torque of both Simu A and B were almost the same as that of real data in the middle of support phase as the valgus torque at knee.

Discussion

Nevertheless, the definition of judgement methods by human eyes on race walking in competition rules of the Athletics[1] and the advocation never to adopt the movie or electric devices in the future,[29] World Athletics has started to discuss the development and adoption of electric judgment system for race walking.[13] The result of the step analysis for Real data of the present study revealed that both the flight time and the flight length were significantly related the walking speed as men's event.[2] Although Hoga-Miura *et al.*[11, 30] has rejected the advantage with the flight time and length, which were related to walking speeds, with multiple regression analyses both for men's and women's event of world elite race walkers, the data of the present study showed that the current judgment system allows the flight phase which occurs along with the walking speed. However, the research about the walking motion so as not to occur flight time would enhance the quality of race walking[2] and help athletes and coaches to achieve higher result in competitions in the future.

Step analysis of the simulated data

Although the walking speed was significantly reduced in the simulated data of men's event with the shortened support length,[2] the results of the step analysis on the simulated data indicate that both Simu A and B were not advantageous to obtain large walking speed and to enhance performance of race walking. Although the flight time of real data for women's event of the present study was almost same as that for men's event,[2] the flight length for women's event was smaller than that of men's event. Then, the reduction in flight length might influence the walking speed of Simu A in the present study less than that in men's event.

Torque analysis of the simulated data

In the previous studies about the mechanical works and powers of male elite race walkers,[10, 27, 31-33] the extensors' and flexors' torque both at hip and knee were related to the walking speed. Also, as Hoga-Miura *et al.*[2] has mentioned the joint torques exerted by muscles of lower extremities, joint torques about knee and hip in sagittal plane were reported to be important to obtain large walking speed.[11, 27, 30, 31, 34] In Simu A, the time of recovery phase was shortened from real data. Reduce of the time of recovery phase may result in the large angular displacement in unit time. Then, because the angular acceleration increased in Simu A rather than real data, knee extensors' torque and hip flexors' torque in the first half of the recovery phase and the knee flexors' torque and the hip extensors' torque in the second half of the recovery phase in Simu A might increase. Large joint torque at knee and hip of recovery leg may result in the large acceleration of the recovery thigh, shank, and foot. These large accelerations may influence the knee flexors' and hip extensors' torque in the first half of the support phase and the knee extensors' and hip flexors' torque in the second half of the support phase through large ground reaction force which was enhanced with the acceleration of the recovery leg. In Simu B, ranges of motion for the recovery leg and the support leg were expanded from real data. Although increase of range of motion may result in the large angular displacement in unit time, the joint torques at knee and hip in the first recovery phase of Simu B which help to swing the recovery leg forward were smaller than those of real data. However, the larger plantar flexors' torque at ankle and the extensors' torque at knee for Simu B seems to help large forward swing velocity of recovery leg just after toe-off. The changes of joint torque in the simulated data from real data of the present study were consistent to the data of the male elite walker without flight phase.[35] In the data of Hoga,[35] walkers whose flight phase could not be recognized exerted abrupt large hip abductors' torque in the middle of support phase both in official race and experiment. Although those torques of simulated data in the simulation of male athlete were different from real data,[2] the data of the present study both in Simu A and B was almost same as that of real data. The support hip position of the walkers without flight phase in the frontal plane was lateral rather than another walker with flight phase.[35] However, the simulation was conducted only in the sagittal plane in the present study. Simulation in the frontal plane was not conducted in this study. Then the simulation procedure may influence the joint torque in the frontal plane through vertical ground reaction force with the simulation onto the recovery leg. However, the simulation in the present study may not influence the vertical acceleration of the recovery leg.

Conclusions

The present study aimed to obtain the walking motion without flight phase for female athletes. From three-dimensional coordinate data of athlete in the real official competition, virtual three-dimensional coordinate data was calculated, and the kinetics was obtained with inversed dynamics. In the result of the simulation which directed to reduce the recovery phase time, the walking speed does not reduce. However, the larger joint torque at hip and knee exerted rather than the real data. In the result of the simulation which directed to expand the support phase time, the walking speed was same as the real data. The large joint torques at ankle and knee in the final phase of the support phase exerted rather than the real data and helps large forward swing velocity in the early recovery phase with lower knee extensors' and hip flexors' torque rather than real data.

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