Abstract. [Purpose] Amputee soccer is a game for individuals with amputations. Players use losfstrand crutches to move around the field and kick the ball. Scoring quick goals during a match requires players to have maximum running skills. Notably, a few parameters affect the running speed in players; however, no study has reported the biomechanical analysis of running in amputee soccer. Thus study aimed to analyze the biomechanics of single-leg running using losfstrand crutches in 12 healthy adult males (6 with prior amputee soccer experience and 6 without such experience). [Participants and Methods] The kinematics of the lower limb and the pelvis, the ground reaction force, and skill in using the crutches were evaluated using 3 dimensional motion analysis combined with 8 force plates. Lower leg amputation was simulated in all participants by maintaining the non-dominant knee in a position of maximum flexion using an elastic band. [Results] Significant differences were observed between experienced and non-experienced participants with regard to the angle of the pelvis and the crutch stance phase. Specifically, higher running speed was associated with an increased forward tilt of the pelvis and a shorter crutch stance phase. [Conclusion] These findings will be useful to improve the running speed of amputee soccer players.

Key words: Disability, Kinematics, Lower limb amputation

INTRODUCTION

Amputee soccer is gaining popularity worldwide among individuals with disabilities\(^1\). Regular physical activity and/or participation in sports reduces the risk of lifestyle-related diseases, which are prevalent among individuals with disabilities, and has positive effects in improving balance, muscle strength, physical fitness, and overall quality of life\(^2,\,3\). Amputee soccer is a variation of conventional soccer in which all the outfield players have a lower limb amputation and the goalkeepers have two legs but with amputation of one of the upper limbs\(^4,\,5\). The outfield players move around the pitch with the use of losfstrand crutches and kick the ball only with their remaining foot. As players use losfstrand crutches to move around the field, the skill in using these crutches will influence their running speed and ball-kicking performance.

Maximum running speed and acceleration along a straight line are fundamental skills in soccer\(^6,\,7\). The biomechanics of running have been described in several previous studies, including an evaluation of the factors at the level of limb kinematics, ground reaction force, and running parameters (including stride length and frequency), which are associated with improvement in running speed\(^8,\,10\). As an example, it is known that stride length and foot stride frequency increase as running speed increases\(^11\). With regard to amputee running, studies have reported on the physical demands of amputee soccer,
including anaerobic performance, muscle strength, sprint performance, balance, and locomotor capacity\textsuperscript{12, 13}. However, the biomechanics of single-leg running with losfstrand crutches in amputee soccer have not been evaluated. This information is required to optimize running performance among amputee soccer players. Therefore, the purpose of this study was to analyze the biomechanics of single-leg running with losfstrand crutches in amputee soccer, and to clarify the relationship between kinematic data, obtained using 3-dimensional (3-D) motion capture, and running speed. We hypothesized that physical movements and crutch operation skills affects running speed in single-leg running with losfstrand crutches.

**PARTICIPANTS AND METHODS**

Twelve physically active male participants were recruited into our study, with 6 having at least 1–3 months of prior experience playing amputee soccer and running with losfstrand crutches, whereas the other 6 participants having no such prior experience. Relevant characteristics of our study group are summarized in Table 1. Their body weight was measured using analog health meter (TANIT, JP). As the length of the stump influences the location of the body center of mass (COM) and, hence, the biomechanics of running, our participants were not amputees. All participants provided informed consent, and the study protocol was approved by the Ethics Committee of the Graduate School of Health Sciences, Hiroshima University (ID: 1525).

Various single-leg running styles are possible using losfstrand crutches in amputee soccer. This study investigated a basic style of running with losfstrand crutches, which is 1:2 ratio of crutch-to-foot contact during a running cycle. However, the cycle of single-leg running with losfstrand crutches has not been defined. The cycle of single-leg running is initiated by crutch contact with the ground and ends with the subsequent ground contact of the crutches. For our analysis, we subdivided the running cycle into the following 4 phases (Fig. 2):

- **Crutch stance phase**, defined as the time the crutch is in contact with the ground to the time the foot is in contact with the ground; first stance phase, which extended from crutch stance to lift-off of the foot from the ground; jumping phase, defined as neither the foot nor the crutches are in contact with the ground; and the second stance phase, defined as the contact of the foot and crutches with the ground following the jumping phase.

A pair of standard aluminium losfstrand crutches of length adjustable type was used in this study. The length of crutches was adjusted for each participant, which it feels easy to use in the running. Prior to data collection, participants were provided with sufficient practice to become accustomed to the demands of single-leg running with losfstrand crutches. To simulate a lower limb amputation, the non-dominant knee was maintained in a position of full flexion, using an elastic bandage (100 mm × 9.1 m; Fig. 1).

3-D kinematic data were recorded using a 16-camera Vicon motion capture system (Vicon Motion Systems, UK; sampling rate, 100 Hz), combined with 8 (2 rows × 4 columns; 80 cm × 240 cm) force plates (AMTI, USA; sampling rate, 1,000 Hz) to record ground reaction forces. Crutch and foot strike was identified as occurring at the first frame for which ground reaction rate, 100 Hz), combined with 8 (2 rows × 4 columns; 80 cm × 240 cm) force plates (AMTI, USA; sampling rate, 1,000 Hz) to record ground reaction forces. Crutch and foot strike was identified as occurring at the first frame for which ground reaction forces of 10 N threshold\textsuperscript{14}. Infrared reflective markers were applied to the participants using the plug-in gait full body model. An additional marker was placed on the distal point of each crutch\textsuperscript{15}. The data of markers were processed using the Vicon Nexus 1.8.5 software (Vicon Motion Systems, UK).

For the experimental task, participants were asked to 10 m straight line run, which included the start and stop, at their maximum speed. The position of the start unified with the crutches grounded to the front of body and started at an arbitrary timing. And Trials were repeated, with sufficient rest between sets, to obtain three trials with successful foot contact on a force plate.

For the analysis, each running cycle was time normalized to 100% for between-participant comparison, and the averaged value across the three trials was used for the analysis. The horizontal speed over one running cycle was calculated as the mean velocity of COM, calculated as the COM distance divided by time. The following sagittal plane kinematic variables were calculated: trunk tilt, pelvic tilt, flexion angle of the hip, knee and ankle angle at first stance, and total joint excursion (defined as the total displacement of the joint over the period from crutch stance to the second stance).

Ground reaction force data were recorded only during the first stance, due to the space limitation of the laboratory, which made it difficult to measure forces over an entire running cycle, from crutch stance to second stance. Again, the force profile was normalized to 100%, and magnitudes normalized to body weight for between-participant comparison. Moreover, we also analyzed horizontal ground reaction force (hGRF) and vertical ground reaction force (vGRF).

| Table 1. Baseline characteristics of participants |
|-----------------------------------------------|
|                                              |
| **Age (years)**                              |
| Total (n=12)                                 |
| Experienced (n=6)                            |
| Non-experienced (n=6)                        |
| p-value                                      |
| 20.8 ± 1.0                                   |
| 20.8 ± 1.2                                   |
| 20.7 ± 1.0                                   |
| 0.80                                         |
| 172.6 ± 5.3                                  |
| 173.5 ± 5.4                                  |
| 171.7 ± 6.0                                  |
| 0.59                                         |
| 62.6 ± 6.2                                   |
| 63.5 ± 5.0                                   |
| 61.7 ± 8.1                                   |
| 0.65                                         |
| 21.0 ± 1.1                                   |
| 21.1 ± 0.8                                   |
| 20.8 ± 1.5                                   |
| 0.75                                         |

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Stride length was calculated from first crutch contact to heel contact, using the marker on the distal crutch and a marker on the foot as a reference. Crutch stride length was defined as the distance between the distal marker of the crutch and to the heel marker (crutch-to-heel; C-H stride) and to the toe marker (crutch-to-toe; C-T stride). The time-point of crutch contact (from initial crutch contact and crutch lift-off) was defined as the time over which the vertical movement of the crutch did not exceed 1 mm.

The mean (standard deviation) of all measured values was calculated, and the normality of the distribution of the data was evaluated using the Kolmogorov-Smirnov Test. Pearson’s correlation analysis was used to evaluate the association between each measured variable and running speed. Differences between the experienced and non-experienced groups were evaluated using an independent t-test. All statistical analyses were performed using EZR ver. 1.35 with a p-value <0.05 considered significant. The confidence interval (95% CI) was calculated to evaluate the magnitude of difference in measured variables between-groups and for running speed.

RESULTS

At first, each data indicated normality. Baseline characteristics were comparable between the two groups (Table 1). The normalized running cycle was as follows: crutch stance phase, 33.3% ± 3.4%; first stance phase, 21.7 ± 1.8%; jumping phase, 22.4% ± 2.0%; and second stance phase, 22.6% ± 4.0%. The average running speed was 3.1 m/s ± 0.3 m/s (range, 2.5 m/s to 3.7 m/s). Crutch stance phase comprised the longest proportion of the running cycle, with the other 3 phases being approximately of the same duration.

Running speed was significantly greater in the experienced group than in the non-experienced group (Table 2), with the pelvic tilt angle and crutch stance time being significantly correlated to running speed (Table 3). Greater running speed was associated with a greater anterior tilt of the pelvis and shorter crutch stance time. The between-group difference in pelvic tilt and crutch stance time was significant (Table 2).
DISCUSSION

To our knowledge, this is the first study to have described the biomechanics of single-leg running with lofstrand crutches within the context of amputee soccer. The finding of this study indicated that there was a significant correlation between running speed and an increase in pelvic tilt angle and a decrease in crutch stance time. Moreover, these two variables specifically differentiated the experienced group from the non-experienced group. In swing-through gait using crutches, the stance phase accounts for approximately 55% of the gait cycle (range, 52% to 57%), with the swing phase accounting for 45% (range, 43% to 48%), at a walking speed of 0.73 m/s. A comparison of the use of lofstrand crutches and prosthetic walking, at a speed of 1.4 m/s reported an average crutch stance phase of 58%, with a crutch swing phase of 42% among amputee soccer players.

In our study, the average crutch stance phase was 33% and the duration of the crutch not being in contact with the ground accounted for 67% in a running cycle at a running speed of 3.1 m/s (range 2.5 to 3.7 m/s). Therefore, as running speed increases, the crutch stance phase decreases. Although a previous study reported an increase in both pelvic and trunk tilt with increasing running speed, only the pelvic tilt was significantly correlated to running speed in our study. Another previous study about general running showed that the pelvis is tilted forward and the COM falls ahead of body in the acceleration phase. The forward pelvic tilt keep the GRF in a position to allow forward acceleration.

An important component of sprinting is the capacity to produce a large propulsive force over a short contact time. In our study, we did not identify a significant correlation between the hGRF and vGRF components of the ground reaction force and running speed over the first stance, with no difference between the experienced and non-experienced groups. A previous study reported that single-leg jumping ability correlated with single-leg running speed in amputee soccer players. Thus, it was difficult for the participants to obtain sufficient progress force during the first stance phase, and it was not clear if the

| Table 2. Findings on outcome measurement |
|-----------------------------------------|
| Total (n=12) | Experienced (n=6) | Non-experienced (n=6) | p-value |
| Running speed (m/s) | 3.1 ± 0.3 | 3.3 ± 0.2* | 2.9 ± 0.3 | 0.01 |
| Hip angle (°) | 35.7 ± 9.5 | 39.9 ± 9.7 | 31.5 ± 7.9 | 0.13 |
| Knee angle (°) | 42.6 ± 6.2 | 40.6 ± 5.6 | 44.5 ± 6.6 | 0.31 |
| Ankle angle (°) | 24.2 ± 5.2 | 22.9 ± 4.0 | 25.5 ± 6.4 | 0.43 |
| Trunk tilt angle (°) | 33.6 ± 8.1 | 38.1 ± 9.2 | 29.2 ± 3.5 | 0.05 |
| Pelvic tilt angle (°) | 13.3 ± 9.9 | 20.4 ± 8.3* | 6.2 ± 5.0 | <0.01 |
| hGRF (N/kg) | 3.6 ± 0.7 | 3.7 ± 0.6 | 3.4 ± 0.9 | 0.52 |
| vGRF (N/kg) | 27.3 ± 2.9 | 27.4 ± 2.9 | 27.2 ± 3.2 | 0.89 |
| C-T stride length (cm) | 59.0 ± 6.3 | 60.2 ± 5.7 | 57.8 ± 7.1 | 0.53 |
| C-H stride length (cm) | 27.4 ± 8.9 | 28.0 ± 8.7 | 26.9 ± 9.9 | 0.84 |
| Crutch stance time (ms) | 276.1 ± 64.5 | 227.8 ± 39.8* | 324.4 ± 44.2 | <0.01 |
| Crutch non-stance time (ms) | 622.5 ± 45.5 | 642.2 ± 46.3 | 602.8 ± 38.4 | 0.14 |

hGRF: horizontal ground reaction force; vGRF: vertical ground force; *significant p<0.05.

| Table 3. Correlation between running speed and each parameter |
|-------------------------------------------------------------|
| r | p-value | 95% CI |
|-----------------|---------|--------|
| Hip angle (°) | −0.006 | 0.99 | −0.58 to 0.57 |
| Knee angle (°) | −0.27 | 0.40 | −0.73 to 0.36 |
| Ankle angle (°) | −0.37 | 0.24 | −0.78 to 0.26 |
| Trunk tilt angle (°) | 0.56 | 0.059 | −0.023 to 0.86 |
| Pelvic tilt angle (°) | 0.58* | 0.047 | 0.012 to 0.87 |
| hGRF (N/kg) | 0.35 | 0.27 | −0.29 to 0.77 |
| vGRF (N/kg) | 0.32 | 0.31 | −0.31 to 0.75 |
| C-T stride length (cm) | 0.45 | 0.14 | −0.16 to 0.82 |
| C-H stride length (cm) | 0.12 | 0.72 | −0.49 to 0.65 |
| Crutch stance time (ms) | −0.87* | <0.01 | −0.96 to −0.59 |
| Crutch non-stance time (ms) | −0.18 | 0.57 | −0.44 to 0.69 |

hGRF: horizontal ground reaction force; vGRF: vertical ground force; *significant p<0.05.
ground reaction force in the foot contact phase influenced forward propulsion.

This study provides a new finding of a shortening of the crutch stance phase as the speed of running increased. In general running, speed is determined by stride length and frequency. Wells reported that as the speed of progression in crutch gait increases from 0.43 to 0.98 m/s, the stride length increases from 0.75 to 1.2 m. In our study, however, there was no significant relationship between running speed and crutch stride length, defined as the distance from crutch to foot, in either the experienced or the non-experienced group. Based on our results, teaching athletes to shorten crutch stance time would be important to increase their running single-leg running speed.

The limitations of our study need to be acknowledged. Foremost was the small sample size. Moreover, due to the limited size of our testing area, with only about 15 m of running space available, the maximum running speed could not be attained. A previous study reported that 0–15 m was the acceleration phase in a 30 m sprint. In addition, there were only 8 (2 rows × 4 columns) force plates, which made it difficult to consistently capture the ground reaction force during the whole cycle of single-leg running with lofstrand crutches. It is important to note that we simulated lower leg amputation to control for effects of different stump lengths on the location of the COM. Thus, future studies are required to fully characterize single-leg running with lofstrand crutches among amputee soccer players with lower limb amputations.

Conflict of interest

There are no conflicts of interest in this study.

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