Available online at www.sciencedirect.com
ScienceDirect
Procedia Engineering 112 (2015) 449 – 454
Procedia Engineering
www.elsevier.com/locate/procedia

7th Asia-Pacific Congress on Sports Technology, APCST 2015

Kinematics of elite unilateral below-elbow amputee treadmill-running - a case study

Franziska Mally [a,b], Stefan Litzenberger[a,b,]*, Anton Sabo [a,b]

[a]University of Applied Sciences Technikum Wien, Department of Sports Engineering & Biomechanics, H¨ochst¨adtplatz 6, A-1200 Vienna, Austria
[b]RMIT University, School of Aerospace and Manufacturing Engineering, PO Box 71, Bundoora VIC 3083, Melbourne, Australia

Abstract

Although scientists have already shown interest in the contribution of upper body movements (e.g. arm swing) to the biomechanics of running and walking, no study so far has investigated the influence of upper limb impairment on running or walking, respectively. Nevertheless, the International Paralympic Committee partly bases its classification system for athletes with unilateral through or above wrist impairment on research done with able-bodied subjects nearly three decades ago. Hence, in this study a high caliber male middle distance runner (age: 26 yrs., height: 183 cm, weight: 67 kg; personal bests: 400m: 0:48.45 min, 800m: 1:50.92 min, multiple medal winner at Paralympic Games, currently being classified as a T47 due to a missing right forearm from just a few centimeters below the elbow on) performed several trials without and with additional weight (0.5kg, cu worn just proximal to the elbow joint) on the impaired arm at running speeds of 12, 16, 20 and 24 km/h on a treadmill. Concurrently a full-body motion analysis using a ten infrared camera Vicon motion analysis system was performed. Stance phase duration, arm swing velocity and angle between hip and shoulder axis were analyzed. It could be shown, that without weight stance phase duration for the right leg was highly significantly (alpha = 95%) longer at all running speeds than for the left leg, that arm swing velocity was faster for the right (impaired) arm. The hip-shoulder angle, however, did not show any significant difference for right or left side. Running with additional weight changed stance phase duration differences between left and right (not significantly), reduced the velocity differences between the impaired and sound limb (not significantly) and changed the rotation between hip and shoulder (significantly). The results of this study indicate that a missing upper limb affects running kinematics more clearly as hitherto assumed.

Peer-review under responsibility of the the School of Aerospace, Mechanical and Manufacturing Engineering, RMIT University

Keywords: running, kinematics, arm swing, upper body rotation, upper limb impairment

* Corresponding author. Tel.: +43 1 333 40 77 377; fax: +43 1 333 40 77 99377.
E-mail address: litzenberger@technikum-wien.at

Peer-review under responsibility of the the School of Aerospace, Mechanical and Manufacturing Engineering, RMIT University
doi:10.1016/j.proeng.2015.07.223

1. Introduction

In 1987 Hinrichs et al. published a major two-sequence article about Upper Extremity Function in Running in the International Journal of Sport Biomechanics [1, 2]. Ever since then scientists have strived to investigate the contribution of arms and upper body to the motion of running a movement supposedly driven by the function of the lower body.

Several studies by Hinrichs et al. [1, 2, 3] investigated the arms contribution to vertical and horizontal propulsion (”lift” and ”drive”) and to angular momentum about all three orthogonal axes of the body. Al- though the legs were found to have the biggest influence on the lift, also the arms showed a positive contribution to vertical propulsion of the body. Hinrichs et al. [1] called this contribution to be small (generally between 5-10%), but potentially important. Furthermore the contribution of the arms was found to increase with increasing running speed. Because the arms movements oppose each other in anteroposterior direction the two arms nearly cancel each other out. However, the relative momentum of each single arm in ante posterior direction was found to be drastically higher than the combined relative momentum of both arms. Lees and Barton [4] confirmed Hinrichs' findings by applying a slightly different algorithm leading to similar overall results.

A number of other studies investigated the influence of different loads on arms and legs [5] and different ways of manipulating arm swing, such as restricting or suppressing arm swing while running and walking [6, 7, 8, 9, 10]. However, all those and further similar studies so far have entirely been conducted with able- bodied, recreational subjects. No published study could be found performing comparable measurements with impaired athletes.

Nevertheless, based on studies such as the one by Hinrichs [3] the International Paralympic Committee (IPC) in 2010 published its final report of Stage 1 of the ”IPC Athletics Classification Project for Physical Impairments” [11]. The report includes a change of classification affecting athletes with unilateral through or above wrist impairment competing in short- and middle distance running events, splitting up the former class of T46 into a new version of T46 and an additional class T47. These changes have been effective since after London 2012.

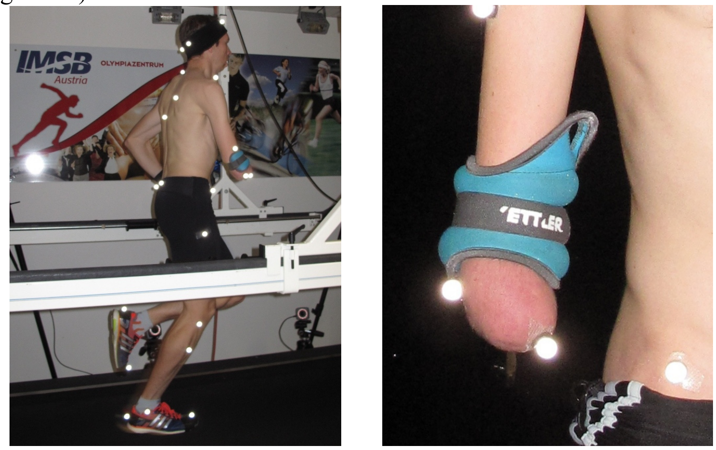
For this study it was hypothesized that compensation of missing an upper limb while running results in a higher arm swing velocity of the impaired arm than of the sound arm (H1). This compensation further leads to differences in stance phase (SP) duration whereas the SP is longer on the side of the impaired arm, because swing duration of the impaired arm is shorter (H2). Due to the hypothesized compensation methods no difference can be seen in upper body rotation (angle between hip and shoulder axes) (H3). Finally, additional weight on the impaired arm is thought to reduce differences between parameters for the sound and impaired side (H4).

2. Methods

An internationally highly successful male Paralympic athlete (age: 26 yrs., height: 183 cm, weight: 67 kg) with personal bests of 0:48.45 min and 1:50.92 min for the 400m and 800m respectively acted as a subject in this study. The subject is currently being classified as a T47 (as of European Championships 2014) athlete by the IPC due to missing his right lower arm from just a few centimeters below the elbow on. He was informed of the purpose and the methods of the study and gave his consent to participating.

Measurements took place at the IMSB-Austria (Maria Enzersdorf, AUT). The subject performed four runs on a h/p/cosmos saturn 2.0 treadmill (h/p/cosmos sports & medical gmbh, GER) at 12 km/h, 16 km/h, 20 km/h and 24 km/h, roughly covering the range between warm-up- and close-to-racing-speed. Addition- ally to that the subject performed another four runs at the same speed levels, with wearing a weighted cu (0.5 kg, Heinz Kettler GmbH und Co. KG, GER) around the right elbow, supposedly making up for the presumably missing mass due to the limb deficiency (Figure 1b).

Twenty seconds of each run were captured using a 3D Vicon Nexus motion analysis system (Vicon Motion Systems, Oxford, UK) with ten Vicon Bonita infrared-cameras, filming with a framerate of 100 Hz. Spher- icalretro-reflective markers with a diameter of 16 mm were attached to the subject on several anatomical landmarks. According to the Vicon PlugInGait FullBody markerset markers were placed on the left and right fore- and backhead, C7 and T10 vertebrae, right scapula, sternum and clavicle, left and right shoulder, upper arm and lateral epicondyle, left forearm, ulnar and radial styloid and second metacarpal, left and right anterior and posterior iliac spine, left and right thigh, lateral femoral condyle, tibia, lateral malleolus, second toe and calcaneus (Figure 1a). The marker set of the right lower arm was adapted due to the missing limb. Hence below the elbow only one marker was placed on the stump (Figure 1b).



(a) (b)

Fig. 1. (a) Subject with equipped with markers running on the treadmill, (b) subject's right arm (impaired limb) with markers and additional cu .

To make a comparison and statistical evaluation of different running speeds possible, a step detection algo- rithm was applied to the data, single steps were separated and time normalized to 100% of the gait cycle. The indices obtained were used to time-normalize the data and the following parameters were calculated for every single step:

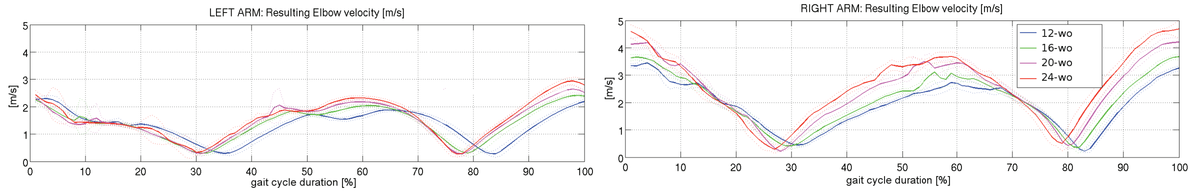
Resulting Arm Swing Velocity for left and right arm were obtained by calculating the resulting velocity of the left and right elbow, respectively.

Stance Phase (SP) Duration is defined as the time between initial contact of the foot and toe o of the same foot. These times were calculated for left and right foot and expressed in percent of the gait cycle (i.e. initial contact of one foot until next initial contact of the same foot). For statistical evaluation the arithmetic mean, the median value, the 25th - 75th, the 9th - 91th percentile, the maximum and minimum value were calculated for each trial and displayed as a boxplot.

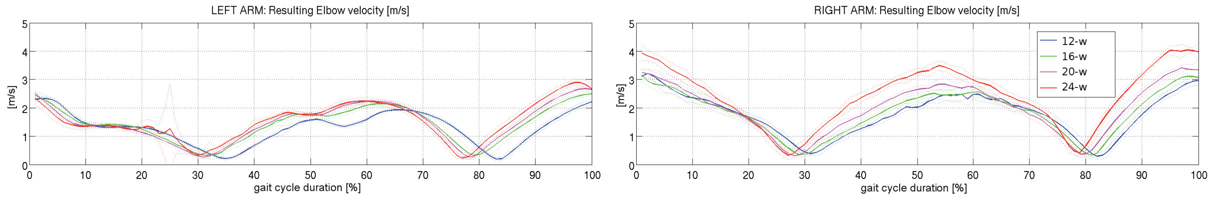
Angle Between Hip- and Shoulder-Axes was calculated using the markers on both the acromia and the four hip markers. The hip axis was formed by calculating the mid-point between anterior and posterior spinae iliacae of the left and right hip and calculating the vector between these two points. To obtain the angle between hip and shoulder the projected angle between the two vectors on the transversal plane was calculated. For displaying the data the mean value for all steps and standard deviations were calculated and plotted over the gait cycle. Furthermore maxima, minima and their occurrence in the gait cycle were recorded.

3. Results

The resulting arm swing velocities were plotted (Figure 2) related to the initial contact of the contralat- eral foot (left foot for the right arm and vice versa). It reveals that, both with and without additional weight, for the left arm (sound arm) the arm velocities are very similar throughout all running velocities, only the 12 km/h data shows slower arm movement than the other trials. For the right arm (impaired arm), however, the results are very different. The velocity is far higher than for the left arm and it can clearly be seen that the arm velocity increases with increasing running velocity. Furthermore there is a slight shift of the velocity profile for the 12 km/h measurements for both arms which is due to the different SP duration. Whereas this is only minimal for the sound limb, these differences are larger for the impaired limb. It must be noted though that for both arms throughout all trials the standard deviation is very small indicating a high reproducibility of the movement.

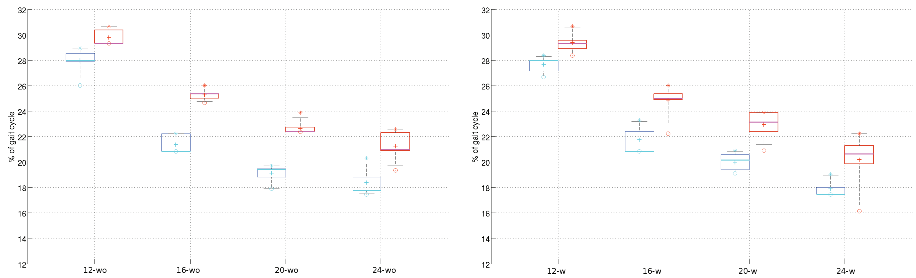


(a) without weight



(b) with additional weight

Fig. 2. Velocity of the elbow marker of left arm (left) and right arm (right), (a) without and (b) with additional weight for four different speeds (12 km/h, 16 km/h, 20 km/h, 24 km/h) plotted over gait cycle duration [% of gait cycle], dotted: 1 SD. It can be seen that v is clearly higher for the right arm, that there is a time-shift for different running speeds and that asynchronous movement between right and left arm occurs.



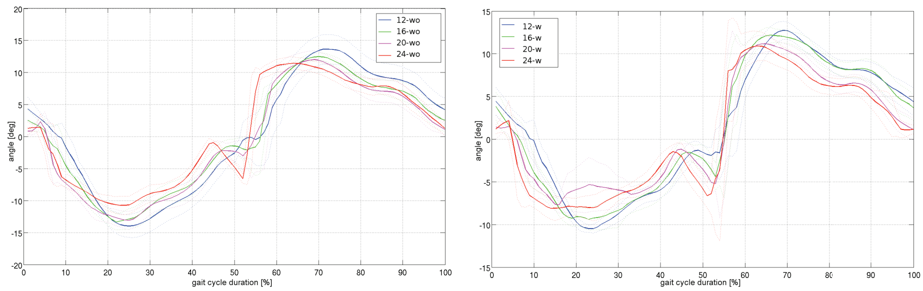
(a) without weight (b) with additional weight

Fig. 3. Duration of the stance phase in % of the gait cycle for four different speeds (12 km/h, 16 km/h, 20 km/h, 24 km/h) (a) without and (b) with using additional weight on the impaired limb. Red: right foot, blue: left foot. Bold line: median, cross: mean, box: 25th 75th percentile, whisker: 9th 91st percentile, *: maximum, : minimum.

SP duration was calculated for all speeds for running without and for running with additional weight. In Figure 3it can be observed that - as was expected - the SP duration decreases with increasing running speed. Furthermore there are massive differences for the SP duration between the left foot and the right foot. The 9thand 91st percentiles do not differ greatly from the mean and median values for all velocities except the fastest (24 km/h) where more variation can be seen. The 25th and 75th percentile are much closer to the median and mean values, only for the left foot at a running speed of 16 km/h almost all values lie within the 25th and 75thpercentile. Table 1 gives the p-values showing that differences between SP duration of the left and right foot are highly significant for all four running speeds. There was, however, no statistically significant difference found for running with or without additional weight except for the left foot while running at 20 km/h ($p < 0:05$).

Table 1. p-value for differences between left and right stance phase duration during running on a treadmill without weight and with additional weight. bold: highly significant (p < 0:01), italic: significant (p < 0:05), plain*: no significant (p 0:05) differences.

	12 km/h	16 km/h	20 km/h	24 km/h
without weight	0.0002	1E-009	5E-009	1E-005
with additional weight	0.0001	7E-006	4E-006	0.0099



(a) without weight (b) with additional weight

Fig. 4. Angle between hip and shoulder axis for four different speeds (12 km/h, 16 km/h, 20 km/h, 24 km/h) (a) without additional weight and (b) using additional weight on the impaired limb plotted over gait cycle duration [% of gait cycle], dotted: 1 SD. Positive values: right shoulder in front, negative values, left shoulder in front (0% signify initial contact with the right foot).

Results for the angle between the hip and shoulder axis are displayed in Figure 4, which shows that the character and the amplitude of the movements are rather similar for all given velocities.

The maximum and minimum angles are reached at the lowest velocity (12 km/h) and decrease with increasing running speeds, meaning less rotation between hip and shoulder at higher running speeds. A noticeable peak in the data for the 24 km/h trial at around 50% of the gait cycle can be seen. This, however, is caused by marker tracking inaccuracy due to marker occlusion at the given point in the gait cycle and should not be considered a movement.

A students t-test revealed that significant differences between rotation directions (i.e. left and right shoulder in front) could be found with additional weight which were highly significant (p < 0:01) for 12 km/h, 16 km/h and 20 km/h and significant for 24 km/h (p < 0:05) These results indicate that the additional weight on the impaired limb do not influence the shoulder-hiprotation for the shoulder ipsilateral to the impaired limb (in this case right) in front but highly influence the rotational behavior when the shoulder contralateral to the impaired limb (in this case left) is in front.

4. Discussion

It can be assumed that the athlete tries to compensate for the lower mass of the impaired arm by adapting his movement (i. e. increasing arm-swing velocity). A problem, however, is posed by the unknown anthropometric parameters of the amputee athlete. Since the weight of the single limbs cannot be derived from the tables used for able-bodied athletes, the masses of the amputee’s upper limbs cannot be calculated properly. Thus a calculation of momentum is impossible and a conclusion regarding the contribution of the upper limbs to momentum can only be made by analysing GRF and the free moment about the z-axis. A correct calculation of the limbs masses and mass distribution could possibly only be done using a3D-bodyscanner which was not available for this research.

It has been shown, that arm swing velocity for the amputee athlete differs between sound and impaired limb. The velocity is higher for the impaired arm as was expected. This behavior can be seen throughout all tested velocities and it is noteworthy that the changes for higher velocities do affect the sound arm to only a very small account whereas the impaired arm shows strong increase with increasing velocity, showing that swing velocity of the impaired limb is higher and increases with increasing running speeds. These findings strongly support H1.

Furthermore they were supported by calculating the most forward elbow position where it could be shown that this occurs much earlier in the SP for the impaired limb (although differences decrease with increasing running speed) indicating vast differences in the kinematics for left and right side.

For upper body rotation the angle between hip and shoulder axes was taken as an indicator. The data recorded show that there is no significant difference for the rotation between the two sides during treadmill running without additional weight. Hence H3 can be regarded as verified. There is similar rotation on either side.

It could be shown, that H2 is strongly supported by the results. For all four running speeds there is a longer SP in % of GC on the side of the impaired limb (right foot) the differences for left and right are highly significant for all running speeds. The absolute duration of the GC however has been found not to differ for left and right foot. As expected and as reported in numerous studies SP duration decreases with increasing speed. The absolute differences for SP duration, however, do not decrease proportionally, but rather stay constant.

H4 could not be verified in general. Effects on different parameters could be observed.

Whereas for SP duration no highly significant differences were found, for the angle between hip and shoulder highly significant and significant differences could be observed for when the left shoulder is in front for running with and without weight, but no difference was found for when the right shoulder was in front.

While the added weight did hardly affect the velocity of the sound arm, swing velocity of the impaired arm decreased with additional weight. Hence arm swing velocity differences between sound and impaired arm decreased. However, the impaired arm still showed higher velocity though.

References

[1] R. N. Hinrichs, P. R. Cavanagh, K. R. Williams, Upper extremity function in running. Part I: Center of mass and propulsion considerations, International Journal of Sport Biomechanics (3) (1987) 222–241.

[2] R. N. Hinrichs, Upper extremity function in running. II: Angular momentum considerations, International Journal of Sport Biomechanics (3) (1987) 242–263.

[3] R. Hinrichs, Upper Extremity Function in Distance Running, ERIC, 1990, pp. 107–133.

[4] A. Lees, G. Barton, The interpretation of relative momentum data to assess the contribution of the free limbs to the generationof vertical velocity in sports activities, Journal of Sports Sciences 14 (6) (1996) 503–511, pMID: 8981289. arXiv:http://www.tandfonline.com/doi/pdf/10.1080/02640419608727737, doi:10.1080/02640419608727737.

[5] R. Ropret, M. Kukolj, D. Ugarkovic, D. Matavulj, S. Jaric, Effects of arm and leg loading on sprint performance, European Journal of Applied Physiology and Occupational Physiology 77 (6) (1998) 547–550. doi:10.1007/s004210050374.

[6] B. R. Umberger, Effects of suppressing arm swing on kinematics, kinetics, and energetics of human walking, Journal of Biomechanics 41 (11) (2008) 2575 – 2580. doi:http://dx.doi.org/10.1016/j.jbiomech.2008.05.024.

[7] W. Tseh, J. L. Caputo, D. W. Morgan, Influence of gait manipulation on running economy in female distance runners, Journal of sports science & medicine 7 (1) (2008) 91–95.

[8] H. Pontzer, J. H. Holloway, D. A. Raichlen, D. E. Lieberman, et al., Control and function of arm swing in human walking and running, Journal of Experimental Biology 212 (4) (2009) 523–534.

[9] R. Miller, G. Caldwell, R. Van Emmerik, B. Umberger, J. Hamill, Ground reaction forces and lower extremity kinematics when running with suppressed arm swing., Journal of biomechanical engineering 131 (12) (2009) 124502.

[10] C. J. Arellano, R. Kram, The effects of step width and arm swing on energetic cost and lateral balance during running, Journal of Biomechanics 44 (7) (2011) 1291 – 1295. doi:http://dx.doi.org/10.1016/j.jbiomech.2011.01.002.

[11] S. M. Tweedy, IPC athletics classification project for physical impairments: Final report - stage 1, online (2010) [cited October 2013]. URL http://www.paralympic.org/sites/default/files/document/120725114512622_2010_07_16_Stage_1-Classification_Project_Final_report__for_2012_forward.pdf