The TripAnalyser: a wearable system to assess gait and potential tripping

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ABSTRACT

A wearable Arduino system (ESP32s) is described that uses optical time of flight (VL6180X) and inertial measurement unit (SENtral EM7180) sensors to estimate the Minimum Foot Clearance (MFC) of participants during gait. It is envisaged that an affordable wearable device that can acquire kinematic data outside the laboratory over periods of several weeks may find application in falls risk assessment in those with ambulatory disorders including the elderly. A geometric model is presented, and a preliminary trial was conducted with able-bodied subjects to test the correlation and agreement of the device with a Vicon 3D motion capture system, consisting of 12 infrared cameras located at the University of Warwick. The correlation between the device and the gait laboratory data yielded a correlation coefficient of r = 0.88. Agreement was tested using the Bland-Altman plot where the line of equality was within the 95% confidence interval of the mean difference suggesting that the device can be used as an alternative to Vicon for estimating MFC.

KEY WORDS: Elderly, Fall risk, Motion capture systems, Wearable optical systems.

1. Introduction

Tripping is one the most common causes of falling accidents in humans. The cause for a fall ranges from accidentally losing balance during the swing phase of the gait to neurological disorders and ageing. What makes falling an important field to research is that it affects the general populace and most importantly the elderly. One of the most common accidents among the elderly is falling, which results in enormous healthcare costs annually. The Public Health Outcomes Framework reported that between 2013 and 2014, 255,000 people aged 65 and over were admitted to a hospital after a fall-related incident in the UK with the annual healthcare cost for treating these people being at 4.4 billion pounds [1]. The increasingly ageing population will continue to strain the healthcare system. Pin et al. [2] found that patients who had already fallen had negatively associated the fall with social participation, and thus restricted their daily activities. Another study by Liddle et al. [3] conducted in 1995 interviewed 69 elderly people, with an age greater than 65 years old who had suffered a fall, and their carers. They concluded that 25% of the elderly patients had developed a fear of falling and that

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58% of the carers feared that the friend/relative might fall again [3].

One indicator for assessing falls risk is the Minimum Foot Clearance (MFC) which according to Winter [4] is an event during the gait cycle, where the foot travels with the maximum horizontal velocity and the distance between the ground and the base of the foot is the lowest. A typical value of MFC for the healthy elderly (over 60 years old) is around 1.12 cm [4]. Exploring the MFC in a real-world setting can give rise to a better understanding of the cause of falls, and/or strategies to prevent falls in vulnerable populations that will drastically reduce the associated healthcare costs and improve their quality of life. Knowing this value, patients can alter their gait patterns to reduce their likelihood of falling. This can be achieved by either reducing their MFC variability or by increasing the MFC height central tendency [5].

Traditionally, the MFC is measured from kinematic data that are acquired using a gait laboratory. Reflective markers are attached to key locations on the patient's lower body and infrared cameras placed around the laboratory capture the 3-dimensional spatial coordinates of the markers with respect to a global reference system as a time series. The downside of such a laboratory is the cost associated with the formation of one (tens of thousands of pounds) and the limited time patients spend in it. Gait laboratories are usually smooth and level floors and may not accurately represent the everyday surfaces and obstacles that people walk over and occasionally trip on. Furthermore, studies have shown that patients tend to exaggerate their gait patterns when under clinical testing conditions, making the data analysis difficult [6]. On this account, the fall's risks may not be discovered in such an environment, which indicates the need for free range measurement. In the literature there are systems that have been used to collect kinematic data and compare them to a motion capture system [7],[8]. However, the idea to combine both proximity and IMU sensors is novel and provide greater sensitivity to the MFC because foot pose has an important role in the desired measurement. Estimating ground or foot clearance using these devices is usually based on the numerical integration of the vertical acceleration of the foot. However, this approach cannot sense the presence of obstacles or uneven floor surfaces which may present trip hazards and IMU sensors are prone to drift so the need for a Kalman filter to reduce noise is essential, especially when the end goal is to get displacements.

TripAnalyser, the prototype wearable optical system we propose, has the capability to sense the foot's proximity to such obstacles and estimate their distance to the foot. The system uses 2 TOF proximity sensors (VL6180X) one placed on the heel and one in front of the second metatarsal, one IMU (SENtral EM7180) with an in-built Kalman Filter and an ESP32s microcontroller. Preliminary results are presented for the accuracy in estimating MFC in a group of able-bodied subjects walking on an even floor in the Vicon laboratory.

2. Determination of MFC

To make use of the TOF and IMU sensors for estimating the distance between the sole of the shoe and the ground, a trigonometric model that accounts for the data from the sensing elements is essential. The model assumptions are as follows:

1.   The distance measured from the TOF sensor is perpendicular to the ground.

2.   The placement of the IMU and the TOF sensors is parallel to the ground.

3.   The shoe creates a plateau from the metatarsal heads to the toes and throughout the calcaneus bone where the vertical distance from the shoe to the ground remains constant.

For the first assumption, the error introduced to the measurement is

$$Er=|1-\cos(\theta)| \qquad (1)$$

where θ is the angle that the photon detected by the receiver has left the transmitter module. Since it is impossible to know the exact angle of the photon that is detected by the receiver, the theoretical maximum error is calculated instead. The TOF sensor has a field of view of 25 degrees, yet photons that originate from half of the field of view are reflected away from the receiver module thus leaving 12.5 degrees of possible photon detection [10]. From equation (1),
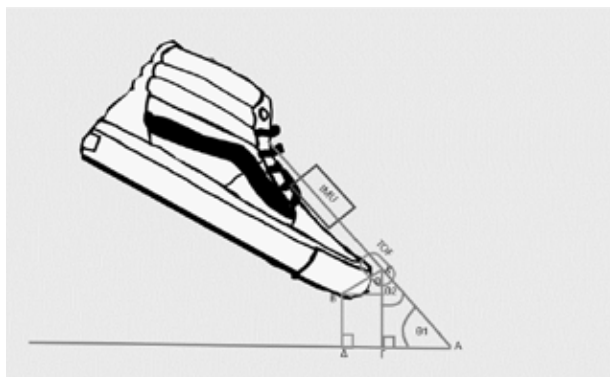
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Figure 1: *Visualisation of the foot during the swing phase. The line AK is the virtual line that provides the inclination of the foot with respect to the ground. The line KB is the initial height of the TOF sensor calculated from the static trial. The line KΓ is the vertical distance from the TOF sensor at that instant and the line BΔ is the vertical distance from the sole of the shoe to the ground.*

Figure 2: *Heel clearance plot of the device and gait laboratory data against the percentage of the swing phase.*

the maximum theoretical error between the hypotenuse and the vertical distance is calculated to be equal to 2.37%. This error corresponds to differences that are less than the accuracy of the TOF sensor for the predicted MFC distances (around 12 mm). The accuracy of the TOF sensor was found to be 2 ± 0.5 mm during initial testing of the device.

The second assumption is crucial to the extraction of the MFC using the trigonometrical model and the placement of the sensor in the experimental procedure is based on it.

The third assumption allows for the placement of the TOF sensor at the front and the back of the shoe without creating significant variations to the model. The plateau idea provides a good estimate of the MFC point because it conceptualises the foot as a one-dimensional object, during the swing phase, which is consistent with approximations done by others in the literature [7],[8],[9].

The proposed model for the extraction of the distance between the sole of the shoe and the ground is shown in Figure 1.

The extraction of the MFC from the toe sensor is going to be analysed, but the same process was followed for the heel sensor as well. In Figure 2, the point K is where the TOF sensor is located, the distance KB is always perpendicular to the line AK as
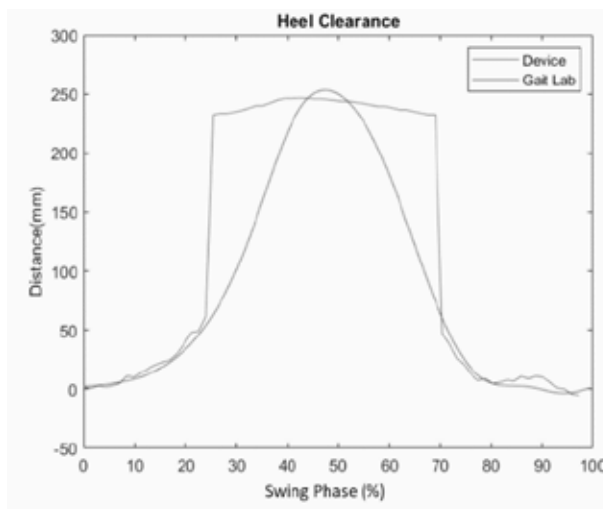
it is the vertical position calculated from a static trial and corresponds to the initial height of the TOF sensor. The distance $K\Gamma$ is the vertical distance measured from the TOF sensor during the dynamic trial and the $B\Delta$ is the instantaneous distance that the foot is clearing the ground. To calculate the MFC, the distance $B\Delta$ needs to be calculated. From the setup, the angle the IMU sensor reports is the same as the angle which the TOF sensor is positioned at with respect to the ground. From the triangle $A\Gamma K$, the angle θ_2 can be calculated as:

$$\theta_2 = 90^\circ - \theta_1 \qquad (2)$$

and because AK is perpendicular to KB

$$\theta + \theta_2 = 90^\circ$$

$$\text{and so } \theta = \theta_1 \qquad (3)$$

Hence the angle reported from the IMU is causally related to the angle the TOF sensor is moved from its original vertical position. Finally, the distance $B\Delta$ can be calculated and it is equal to:

$$B\Delta = K\Gamma - KB*\cos(\theta) \qquad (4)$$

Equation 4 is a generalisation of the standard method used in previous MFC measurement experiments where the value was calculated by subtracting the vertical dynamic trial distance of the TOF sensor, from the mean static trial value [9]. The MFC is then calculated from (4) by finding the localised minimum during the mid-swing phase.
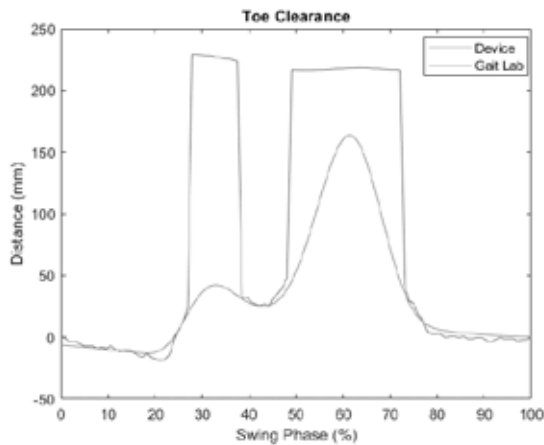
Figure 3: *Toe clearance plot of the device and gait laboratory data against the percentage of the swing phase.*

Figure 4: *Regression plot between device and gait laboratory data.*

3. Results

Ten able-bodied participants were recruited for the study and one static and 5 dynamic trials were performed. From the static trial the distance KB is calculated. This project was granted full approval from the Biomedical and Scientific Research Committee (BSREC) at the University of Warwick (reference number BSREC 53/18-19). The results from the study can be found in Table 1 and the values shown are the average MFC values from the 5 dynamic trials for each participant. The values found under the MFC device column correspond to the data acquired from the developed model only for the toe TOF sensor. The heel clearance data were disregarded because during the instant of MFC the heel is descending towards the ground to prepare for the heel contact portion of the gait cycle. This descending motion does not provide a local minimum value at the MFC point, as shown in Figure 2, and thus was disregarded. The negative regimes shown in Figures 2 and 3 are products of the model. During the initial stage of the swing phase, the toes are pushed downwards to allow the heel to be lifted upwards giving distances lower than the mean static value and resulting in negative portion of Figure 3. Similarly, the heel clearance plot presents a negative region from the moment the heel contacts the ground up until the end of the swing phase. This is attributed to the lower vertical distance of the heel sensor compared to the mean value of the static trial.

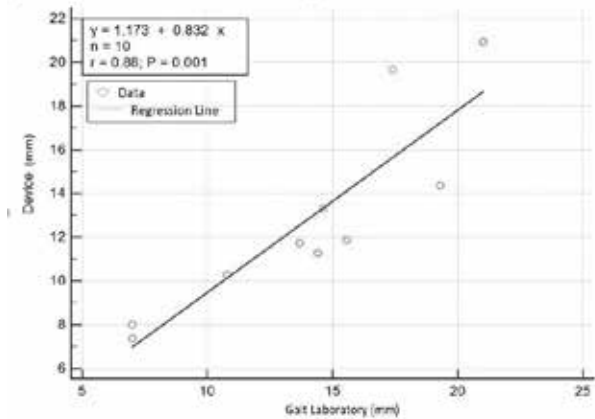For the analysis of the results the MedCalc software was used with the significance level set to 0.05. Linear regression and the Bland-Altman plot were used to test for correlation and agreement between the two methods.

From the linear regression analysis, the coefficient of determination R^2 is 0.7760, which determines the goodness-of-fit and the regression equation where Y is the MFC from the device data and X is the MFC from the gait laboratory data is:

$$Y = a + b \cdot X \qquad (5)$$

with a=1.173 ± 2.3350 mm with a 95% confidence interval (C.I.) of -4.21 to 6.56 mm with a p-value of 0.6290, b=0.8320 ± 0.1580 with a 95% C.I. of 0.476 to 1.1964 with a p-value of 0.0008 and correlation coefficient r = 0.88 with a p-value of 0.001. From the parameters obtained during the analysis, the value of the intercept is not statistically significant because its p-value is greater than 0.05 and there is insufficient evidence in the sample to conclude that a non-zero correlation exists. However, the slope of the equation is statistically significant since its p-value is less than 0.05 suggesting that the null hypothesis can be rejected and that changes in the gait laboratory data are associated with changes in the device data. This effectively shows that there is good correlation between the two methods. This is further supported by the excellent correlation coefficient of the two methods r = 0.88, which is also statistically significant. Figure 4 shows the regression plot.

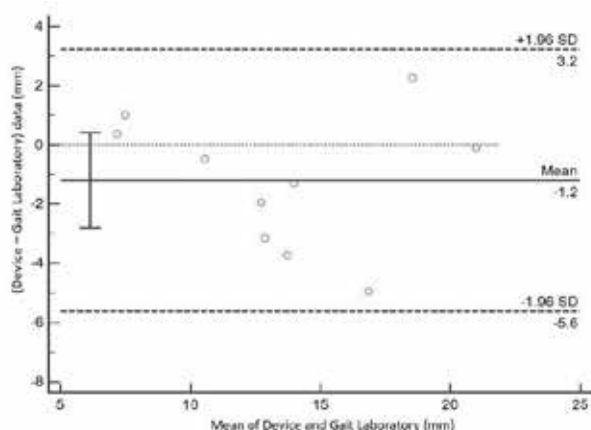To assess agreement between the device and gait

Figure 5: Bland-Altman plot. The dotted line represents the line of equality and the two dashed lines represent the LoA. The straight line represents the mean difference value and the vertical line represents the 95% C.I. of the mean difference.

laboratory, the Bland-Altman plot was used. This graphical representation allows quantification of the agreement between two quantitative methods by constructing the limits of agreement [11]. To use the Bland-Altman plot, the differences between the two methods must follow a Gaussian distribution [11]. To verify this requirement, a Shapiro-Wilk test was used, and the analysis yielded that the difference data are normally distributed with a p-value of 0.9728. This allows the use of the Bland-Altman plot analysis. In Figure 6, the Bland-Altman plot can be seen and in the *Y* axis the difference between the MFC values of the device and gait laboratory data is compared against the *X* axis which has the average of these measurements [11]. This allows for the calculation of the Limits of Agreement (*LoA*) from the value of the mean difference and the standard deviation (*sd*) of the differences according to equation (6).

$$LoA = \pm 1.96 \cdot sd \qquad (6)$$

By constructing the *LoA*, it is expected that 95% of all data points would lie within these regions [11]. Furthermore, the Bland-Altman plot can assess the existence of bias in any of the methods and whether this is significant. This is done by finding the mean difference and its 95% C.I. If the line of equality, where the mean difference is zero, is within the 95% C.I. of the mean difference, then the estimated bias is not sig-

nificant, and the mean difference can be considered as being equal to zero [11].

From Figure 5, the mean difference between the device and gait laboratory data is -1.2 mm with a 95% C.I. of -2.8063 to 0.4194 mm. The yellow dotted line represents the line of equality and the two red dashed lines represent the *LoA*.

The mean difference value suggests that there is a bias of -1.2 mm for the gait laboratory data, meaning that on average the MFC value from the gait laboratory measures 1.2 mm more than the device data. The negative bias is attributed to measurements over 10 mm where a negative trend is apparent from the plot, whereas for measurements below 10 mm the data are closer to each other. However, the estimated bias is not significant because the line of equality is within the 95% C.I of the mean difference value.

4. Discussion

From the analysis of the MFC results it becomes apparent that there is an excellent correlation between the two methods from the regression analysis. Furthermore, it has been established that the device can provide adequate results and it can be used as an alternative to the gait laboratory to extract the MFC. An interesting result is the ambiguity of the intercept from the regression analysis. The slope of the line can be estimated with adequate precision b=0.8320 ± 0.1580 with p-value of 0.0008, but the intercept is not well defined, and its value is not statistically significant. This can be attributed to the placement of the device on the foot for each participant. The intercept is linked to the model's initial conditions (IMU inclination, and static value of the sensor's height) and the ambiguity of its value derives from the different position of the sensors and markers in each participant. To acquire a better statistical value for the intercept, the creation of a standardized placement procedure that incorporates the sensors at predetermined positions, which would be the same for each participant, is necessary to reduce the ambiguity. In both Figures 2 and 3, the plateau portion that can be seen at around 240 mm is attributed to the range limitation of the TOF sensor. The TOF sensor used can sense objects from 0-150 mm and in the configuration used outputs a 255 value when nothing is detected within

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TABLE 1.

Minimum Foot Clearance values from device and gait laboratory data

MFC Device (mm)	Standard Deviation (mm)	MFC Gait Laboratory (mm)	Standard Deviation (mm)
11.87	2.49	15.57	3.62
11.28	3.05	14.42	2.06
7.37	3.10	7.00	2.08
10.30	3.15	10.78	3.60
11.75	2.34	13.70	5.34
8.01	2.89	6.99	3.42
14.37	3.23	19.31	5.42
19.66	8.05	17.39	5.32
20.94	5.32	21.03	4.66
13.33	3.59	14.62	3.80

TABLE 2.

Limits of Agreement from Bland-Altman plot.

Lower LoA (mm)	95% C.I. (mm)	Upper LoA (mm)	95% C.I. (mm)
-5.6126	-8.4680 to -2.7572	3.2257	0.3703 to 6.0811

the user specified time window. For the purpose of this experiment having these regions does not provide any error in estimating MFC since they are not considered in the calculations. The plots were used to visualise the motion of the toes and heel during the swing phase. However, it would be advantageous to have a better representation of the curve, so it is suggested that a different TOF sensor be used. At time of writing this paper, similar TOF sensors (VL53L0X, VL53L1X) have greater maximum range detection but the minimum is 50 mm and 40 mm respectively [12],[13]. A case can be made that for the creation of the device these limitations should be taken into consideration and the newest models should be incorporated; however, we feel that an addon solution has to be as small and unobtrusive as possible especially for the elderly population.

5. Conclusion

This paper introduced a novel device that measured the MFC from participants and compare its values against the gold standard method of a motion capture system. From the statistical analysis performed it is safe to assume that the proposed device can be used as an alternative to the gait laboratory in order to extract the MFC. This is evident from the excellent correlation and agreement between the two methods presented during the analysis of the results. The potential advantages of using the device as an alternative to a motion capture system are the longer periods of recorded data, outside the gait laboratory, over different terrains and floor textures and the relative low price of such a device compared to expensive motion capture laboratory costs. Such a device may provide clinicians with a better estimation of the patient's MFC and MFC variability.

6. Acknowledgments

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