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Lateral loads applied by pedestrians at normal walking velocities

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Abstract. The issue of horizontal loading from pedestrians has received increased attention from bridge designers and researchers in the past decade, primarily due to notable instances of excessive vibration of structures subjected to this form of excitation. Nonetheless there is a scarcity of reliable information on the magnitude and nature of this type of loading. The authors have carried out over 100 walking trials on 27 healthy adult participants walking at normal velocities on a rigid walkway mounted with a force plate. Subject data, pertinent tempro-spatial parameters of gait, walking velocity and pacing frequency are presented for each participant. Additionally, the lateral forces recorded during these tests are presented and analysed. A simplistic force function, based on the fundamental frequency of the applied excitation force, which may approximate the actual load applied by individual pedestrians is proposed. Further, this function is improved by consideration of the lateral force contribution at higher order harmonics of the fundamental frequency and relevant dynamic load factors and phase angles associated with the individual force functions are derived and optimised.

Keywords: Pedestrian, horizontal force, simulation, normal walking, gait, force

1. Introduction

Vibrations induced in structures due to human loading have increasingly become the focus of considerable research in the structural engineering domain in the past decade. This is due primarily to the excessive vibrations experienced by structures such as footbridges, sports stadia, lightweight or long-span floors and staircases when subjected to dynamic loads from pedestrians [25]. Nonetheless, there is a relative dearth of information on the direct nature of pedestrian loading, particularly in the horizontal plane. Attempts have been made to address this through direct measurement of the forces applied from individual footfall force traces in laboratories, employing techniques and equipment traditionally belonging to the biomechanics domain. These individual footfall force traces are commonly referred to as ground reaction forces (GRF's).

1.1. Ground reaction forces

Davis and Kaufman [10] assert that human locomotion (of which walking is one particular type) is fundamentally related to the production of muscle force that either creates or controls the motion. Ground reaction forces constitute an external reflection of these forces. Essentially, ground reaction forces represent the forces applied to a surface by a person walking across the surface and are recorded for each footfall. Measurement of these force traces is normally carried out using a force plate or a force sensor mat. Instrumented treadmills have also been used for this purpose.

Walking imparts forces in three orthogonal planes – longitudinal, lateral and vertical. In biomechanics literature, these three planes are labelled the saggital, medio-lateral and vertical respectively. Bachmann and Ammann [3] reported that the vertical force is of greatest magnitude, followed by the saggital and then the medio-lateral forces. Fig. 1 shows a typical trace of the GRF's recorded in each of the three planes from one strike of a person's foot on a force plate while walking. The peak dynamic component of the vertical load can

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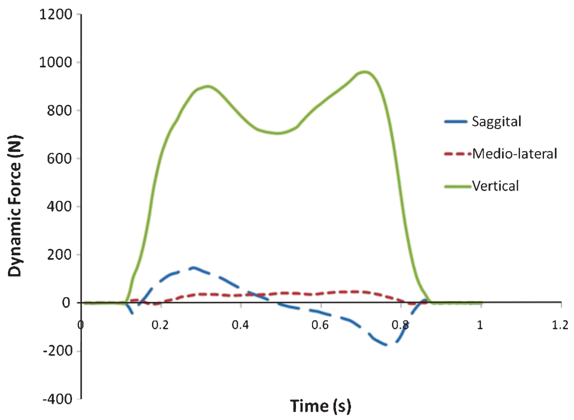

Fig. 1. Typical ground reaction force traces in three orthogonal planes.

be as high as 37% of the person's static weight, while the magnitude of each of the two horizontal components is reported to only be in the range of 3–4% of static weight [3].

In terms of pedestrian loading on footbridges, the saggital plane is not considered to be of consequence as the structure will almost certainly be rather stiff in the direction of walking. The medio-lateral force pattern, generated by successive summation of individual footfall force traces, is of concern when analyzing potentially flexible structures, even though it is of smaller magnitude than the loading applied in either of the other planes. Moreover there appears to be a dearth of reliable published data on the magnitude and nature of this particular loading regime. Zivanovic et al. [32] carried out a comprehensive review of existing data on pedestrian loading and report only two references [3, 4] which provide values for the magnitude of lateral loading in terms of the individual weight of the pedestrian. Further, these two reports vary considerably in their estimation of these values, with the dynamic load factor (the maximum lateral load expressed as a percentage of the static weight) ranging from 3.9% to 10%. Other authors offer values of approximately 4–5% as the ratio of peak medio-lateral force to static weight ([14, 29], cited in [19]). Kirtley [19] also cites Carlsöö et al. [6] who claim that the magnitude of the medio-lateral force increases with step width.

A sequential combination of these individual footfall traces will produce the relevant continuous lateral load pattern exerted by humans walking. The exact nature of this continuous lateral load function will be influenced by both gait parameters and anthropometric data for the pedestrians involved. The primary anthropometric data of concern is the static weight of the person, while the gait parameters which have been asserted to influence the lateral load function are described below.

1.2. Step width

Step width is defined as the distance between the centre lines of the two feet, perpendicular to the plane of walking. Reported values of step width have proven to be quite variable, with standard deviations up to 30%. Further, there is less reported data on this particular spatial parameter than others such as step length. Archbold and Mullarney [2] report a review of current literature, citing references which yield values between 0.09 m and 0.19 m for adults, with no apparent link between subject height and step width. Interestingly, values reported by Cho et al. [8] and Ryu et al. [26] suggest that Korean adults exhibit greater step widths than others reported. Donelan et al. [11] and Bauby and Kuo [5] both linked step width to stride length reporting that the step width was approximately equal to 12% and 13% respectively of the stride length. However, this relationship has not been found by others. Kirtley et al. [19] reported that step width can vary with age and so recommended normalizing the value by dividing it by the pelvic width. They also stated that step width increases with disequilibrium (lack of balance). As previously stated, Calsöö et al. [7] have contended that the magnitude of the lateral load is proportional to step width.

1.3. Foot landing position

Foot landing position is perhaps more commonly referred to as angle of gait, although it has also been termed foot placement angle (FPA) [17] or angular deviation of the foot ([7], cited by [18]). The term generally refers to the angle made between the centerline of the foot and the forward direction of walking, but exact definitions of the foot reference line can vary between authors [30]. Simpson and Jiang [27] defined this reference line as "a line drawn from the mid-point of the posterior aspect of the *calcaneus* to the head of the second *metatarsal*", a definition which will be used here. The same authors also reported tests, which revealed that foot landing position influenced the force applied by the pedestrian. They categorised their test participants into categories of "toe-in", "neutral" and "toe-out" depending on their foot landing position during straight line walking as shown in Fig. 2 and

Fig. 2. Foot landing position (FLP) categories [27].

they claim that toe-out participants exerted significantly greater lateral forces than those in the toe-in category. Values for foot landing position are reported in degrees, with positive representing toe-out and negative representing toe-in. Reported values for foot landing position range between+ 14.3° (toe-out) and −3.8° (toe-in) [27]. This parameter presents the most variability of all of the spatial gait parameters in healthy test subjects. [21] for example, reported mean foot landing positions for two trials each on the left and right foot respectively of 6.73°, 7.32°, 5.01° and 5.02° with accompanying standard deviations of 4.96°, 5.36°, 5.77° and 5.92°. Nonetheless, Kirtley et al. [19] suggest a neutral foot landing position of + 15° i.e. slightly abducted or "toe-out" when measured from the plane of walking, while Taranto et al. [28] measured a mean value of approximately 9°. Chung et al. [9] report a mean neutral value of 13.4° and claim that toe-out participants exerted significantly greater medio-lateral forces than either toe-in or neutral participants in walking trials.

1.4. Pacing frequency

Pacing frequency is the most relevant of the temporal gait parameters in terms of pedestrian loading, particularly where resonant effects on structures are to be considered. It is defined as the inverse of the time taken from the initial contact of the left foot with the ground to the initial contact of the right foot immediately thereafter and corresponds to the rate of application of vertical forces. In biomechanical terms, this parameter is often measured as cadence, which is the number of steps per minute rather than the number per second.

Reported values of normal pacing frequencies indicate that the average pacing rate is between 1.8 Hz and 2.2 Hz. Keogh et al. [17] reviewed 7 references and derived an average pacing frequency of 1.96 Hz, with a standard deviation of 0.21 Hz. Archbold and Mullarney [2] present the results based on a survey of a further 20 sources of information and report a mean value of 1.92 Hz. They also show a slight decreased in pacing frequency with age. Interestingly, results from Oberg et al. [23] indicate that there may be a gender consideration relating to pacing frequency, with women walking at an average of 2.1 Hz and men at an average of 1.98 Hz for normal walking velocity.

The pacing frequency is quite intuitively dependent on the walking velocity, but for the normal range considered here, it has proven to be quite consistent, with published standard deviations of the order of 6%.

2. Lateral load simulation

As previously mentioned, there is relatively little published information on direct simulation of lateral loads from walking humans. A number of approaches have been employed however.

2.1. Single harmonic sine function

Several authors have attempted to define the lateral load pattern as a sinusoidally varying function with a single harmonic, which is a function of the pacing frequency. This approach assumes that the function is perfectly periodic and that the load contribution from alternate footfalls are equal. It also assumes that the use of a single harmonic of the frequency of load application is sufficient to capture the nature and magnitude of the load. It is convenient to note some of the characteristics of such a function at this point.

Firstly, the fundamental frequency of application of lateral walking loads is half the pacing frequency as it is related to successive contact of either the left or right foot with the walking surface. Secondly, the magnitude of the force is assumed to be directly related to the static weight of the pedestrian. The magnitude is thus expressed as a proportion of this static weight through use of a dynamic load factor (DLF). The function can thus be represented as follows:

$$F(t) = L_f\, G \sin(\pi f_s t) \tag{1}$$

where F(t) is the continuous lateral load function, L_f is the dynamic load factor associated with the function, G is the static weight of the pedestrian, f_s is the pacing frequency and t is time. The magnitude of L_f has been reported as ranging from 0.03 [24] to 0.1 [4]. Archbold [1] asserted that the value of L_f may also be influenced by individual temporo-spatial parameters such as foot landing position and not just the static weight of the person. This was used to explain the significant differences in lateral response caused on a lightweight, flexible footbridge by two people of similar weight and height, walking at the same pacing frequency. Erlicher et al. [12] meanwhile demonstrated an increase in the recorded values for lateral force as the pacing velocity increased. The dynamic load factor appears to have increased from approximately 4% while walking at 3.75 km/hr up to approximately 6% while walking at 6.0 km/hr. Ingolfsson et al. [15] calculated a rms value of the lateral load and equated this to 4.1% of the static weight.

2.2. Multiple harmonic sine function

Other authors have attempted to model the lateral force function more accurately by considering several harmonics of the fundamental frequency. The previous assumptions about periodicity and magnitude related to static weight also apply. In this case, the function can be written as a Fourier series as follows:

$$F(t) = \sum_{i=1}^{n} F_{Li}\ \sin(i\pi f_s t + \phi_i) \tag{2}$$

where i is the harmonic number, F_{Li} is the dynamic load factor associated with harmonic i and ϕ_i is the phase angle associated with harmonic i.

Bachmann and Ammann [3] claim that the phase angles can all be treated as zero as only the first harmonic is likely to be of consequence. Moreover, the contributions from harmonics where i is an even number are small. This assumption is further supported by Erlicher et al. [12]. Values for F_{L1} range from 0.039 to 0.1 [3], while values for F_{L3} range from 0.043 [3] to 0.1 [4].

This paper presents the findings from a series of walking trials aimed at determining the nature of the forcing function and values for the dynamic load factors (DLF's) and phase angles associated with lateral loading from pedestrians walking at normal velocities on a rigid surface. Results of the measured gait parameters are presented and a load model is developed for estimating the lateral force generated by a pedestrian on a rigid surface in terms of static weight and pacing frequency. This load model may be of use to practicing engineers and researchers attempting to simulate dynamic lateral loading applied to footbridges from crossing pedestrians in order to estimate the bridge response. Moreover, it may contribute to the development of robust design guidance for the engineering community, which is currently lacking.

3. Experimental programme

The experimental programme reported herein consists of walking trials involving thirteen female and fourteen male healthy adult participants. The participants conducted the walking trials in the laboratory on a specially constructed rigid walkway as described in the following section.

3.1. Participants

Participants were recruited from staff and students at Athlone Institute of Technology, Ireland. All were aged between twenty and forty years. The ethnical composition of the participant sample was predominantly Caucasian with a small proportion being of African and Chinese background. Persons were excluded from participation if they had a history of previous injury with ongoing symptoms, or significant previous injury that would hamper their gait. All participants gave written consent according to the ethical procedures approved by Athlone Institute of Technology and its Research Ethics Committee. Twenty-seven individuals participated in the trials, fourteen male and thirteen female. The sample population was deemed to be reasonably representative of a healthy adult population, as evidenced by analysis of the recorded anthropometric data.

3.2. Anthropometric data

The following parameters were recorded for each test participant prior to the walking trials being carried out: age; height (with and without footwear); weight; right and left leg lengths (measured from the *superior palpable* of the *greater trochanter* to the base of the *lateral malleolus).* A summary of the recorded values is presented in Table 1. The average heights of the participants align very well with reported average male and female heights in Europe. Garcia and Quintana-Domeque [13] for example, reported on the evolution of adult height in Europe. They note that Irish males born

Table 1
Age and anthropometric data for each gender group

Parameter	Male		Female	
	Mean	S.D.	Mean	S.D.
Age (Year)	27.7	5.2	25.0	2.8
Height (m) (with footwear)	1.78	0.06	1.62	0.05
Weight (kg)	82.04	13.50	62.17	14.19
Right leg length (m)	0.86	0.06	0.87	0.05
Left leg length (m)	0.80	0.05	0.80	0.05

between 1970 and 1980 – the majority of the sample population used here – have an average height of 1.77 m, with an overall average from 10 European countries of 1.776 m, compared with mean of this group, which was 1.78 m. The mean reported Irish female height from the same authors was 1.64 m, with a European average of 1.656 m, as opposed to the mean of this group of 1.62 m. The participants were all of or close to healthy body mass index values also, indicating that the sample population accurately represented mean values of adult height and weight.

3.3. Equipment

A rigid walkway was specially constructed to carry out the walking trials. The walkway is 0.9 m wide × 11.0 m long and is constructed from three 50 mm thick laminated fibreboard panels framed with timber battens and cross members at 600 mm centres, which were bolted together longitudinally and placed directly on the laboratory floor. A 500 mm × 500 mm AMTI AccuGait balance platform (force plate) was mounted at the mid-point of the walkway to record the ground reaction forces: the top surface of the force plate was made level with the top surface of the walkway. In the vertical direction, Fz, the force plate has a natural frequency of 150 Hz and a loading capacity of 1334N and the force plate was calibrated prior to the walking trials through measurement of static forces. Three Monitran MTN1800 accelerometers, with a sensitivity of 1.020 V/g@80 Hz, were mounted to the underside of the walkway at approximately one-third span, mid-span, and two-third span respectively.

Data were recorded from the accelerometers through a virtual instrument (VI) developed in National Instruments (NI) LabView 8.5. These data were then used to determine the time interval between consecutive footsteps. Grid paper measuring 3.5 m × 0.6 m and containing a 20 mm × 20 mm grid size was placed over the middle section of the walkway to assist in recording the spatial parameters such as step length, step width and foot landing position from the trials. A schematic layout of the test set-up is shown in Fig. 3.

3.4. Experimental procedure

The participants were asked to wear their regular clothing and comfortable, flat-soled shoes for the walking trials. Prior to the recorded traversing of the walkway, each participant completed a number of 'dummy' runs to ensure they felt comfortable with the process. For these dummy trials and the actual walking trials, the test subjects were requested to walk in a straight line along the length of the walkway at a normal speed, while looking straight ahead – this was aided through using visual targets on the facing walls. Immediately prior to each trial the participant coated the soles of their shoes with blue chalk dust, which aided the recording of the footfall positions and thus measurement of the spatial gait parameters. This procedure has been successfully used by other authors [16, 20, 28, 31]. Each test subject completed a minimum of four recorded trials. Immediately after each trial was completed, the stationary or static step width and foot landing position were also recorded. Figure 5 shows a recorded trial in progress.

The spatial and temporal gait parameters recorded for each trial were step length, stride length, step width, foot landing position, and pacing frequency. Step length is measured as the distance from the heel strike of one foot to the next heel strike of the opposite foot and is measured in the direction of walking. Step width is measured as the distance between the centerlines of consecutive heel strikes and is measured normal to the direction of walking as shown in Fig. 4. This figure also shows the measurement of the foot landing posi-

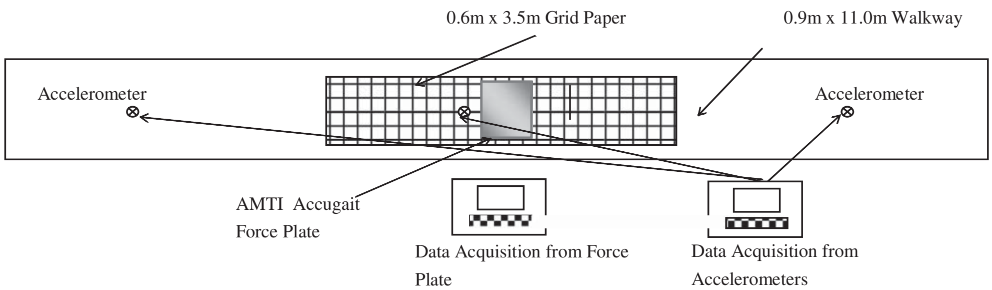

Fig. 3. Schematic representation of walkway and test set-up (not to scale).

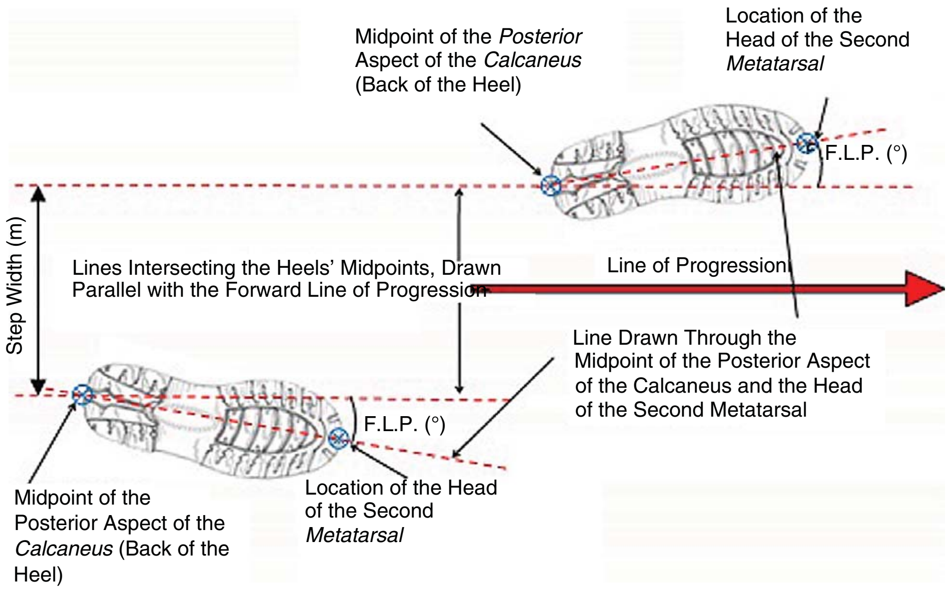

Fig. 4. Measurement of step width and foot landing position.

Fig. 5. Walking trial in progress.

tion, which is defined as the angle made by a line drawn from the centre of the heel, through the head of the second metatarsal and a line drawn parallel to the direction of walking. This is also shown in Fig. 4.

Pacing velocity was determined from the product of pacing frequency and step length. Also, the ground reaction forces (GRF's) in three orthogonal directions were measured for the instance of a footfall striking the force plate. The GRF traces also enabled the determination of the single foot stance support phase in the time domain. The participants completed as many trials as were required to ensure the measurement of at least four complete footfall traces, including a minimum of two from each foot.

4. Results and discussion

4.1. Temporo-spatial parameters

Table 2 presents a summary of the temporo-spatial gait parameters recorded for each participant, along with the population mean values. The mean pacing frequency from the trials was 1.88 Hz, with the mean for males recorded as 1.82 Hz, while the mean for females was higher at 1.92 Hz. However, the mean female step length (0.70 m) was shorter than the mean male value (0.78 m), thus there was only a minor difference between the mean pacing velocities. This difference between genders in terms of pacing frequency is in agreement with the results published by Oberg et al. [23].

The mean values for foot landing position and step width for the entire test population were 6.03° and 0.079 m respectively. Figure 6 illustrates the distribution of these measured parameters. It can be seen that the foot landing position and step width exhibited large scatter in the recorded values as evidenced by the high values for the respective standard deviations. It is notable that these two particular parameters have been

Table 2
Temporo-spatial gait parameters and derived lateral load function characteristics

Test subject	Step length (m)	Pacing frequency (Hz)	Pacing velocity (m/s)	Foot landing Position (degrees)	Step width (m)	L_f, DLF from the single harmonic Sine Function	Multiple harmonic sine function: Dynamic load factors				Phase angles (rads)		
							DLF_1	DLF_3	DLF_5	DLF_7	$\phi_{3(rad)}$	$\phi_{5(rad)}$	$\phi_{7(rad)}$
M1	0.78	1.86	1.43	9.8	0.09	0.075	0.048	–0.030	0.011	–0.008	–1.38	1.21	–2.23
M2	0.78	1.69	1.32	6.2	0.07	0.044	0.034	–0.018	0.005	–0.008	–1.34	1.20	–2.16
M3	0.77	1.80	1.43	5.6	0.12	0.058	0.058	–0.020	0.007	–0.008	–1.39	1.43	–1.99
M4	0.84	1.82	1.53	4.1	0.08	0.047	0.031	–0.018	0.008	–0.004	–1.39	1.10	–2.17
M5	0.78	1.86	1.44	9.8	0.03	0.094	0.037	–0.014	0.013	–0.017	–1.37	1.21	–1.92
M6	0.72	1.64	1.17	4.1	0.10	0.083	0.050	–0.034	0.011	–0.014	–1.33	1.40	–1.90
M7	0.80	1.96	1.63	11.2	0.03	0.081	0.031	–0.025	0.009	–0.014	–1.45	1.44	–1.97
M8	0.79	1.89	1.50	6.9	0.08	0.070	0.044	–0.029	0.009	–0.014	–1.42	1.33	–2.10
M9	0.68	1.90	1.30	14.7	0.07	0.089	0.072	–0.038	0.010	–0.008	–1.41	1.30	–1.96
M10	0.84	1.79	1.50	8.4	0.04	0.050	0.036	–0.021	0.007	0.009	–1.34	0.92	–1.59
M11	0.83	1.82	1.50	4.9	0.10	0.098	0.054	–0.033	0.015	0.013	–1.55	0.94	–1.50
M12	0.85	1.84	1.56	9.5	0.09	0.092	0.045	–0.033	–0.015	–0.012	–1.36	0.37	–1.88
M13	0.77	1.81	1.40	14.8	0.05	0.106	0.048	–0.030	–0.013	–0.012	–1.41	0.20	–1.95
M14	0.74	1.85	1.37	11.0	0.11	0.084	0.080	–0.040	–0.010	–0.008	–1.36	0.27	–2.17
F1	0.65	1.77	1.14	–1.3	0.17	0.039	0.039	–0.004	–0.002	0.002	0.30	–0.22	1.08
F2	0.70	1.88	1.32	8.5	0.05	0.062	0.038	0.016	–0.010	–0.010	–0.36	0.37	0.16
F3	0.80	1.88	1.51	8.3	0.05	0.066	0.057	0.028	–0.011	0.011	–0.62	–0.06	0.53
F4	0.57	1.74	1.00	3.7	0.10	0.063	0.051	0.021	–0.010	0.009	–0.61	–0.15	–1.61
F5	0.73	2.03	1.49	–0.4	0.13	0.085	0.057	–0.034	–0.009	0.011	–1.45	–0.13	–1.61
F6	0.75	1.96	1.46	8.4	0.05	0.069	0.038	–0.023	–0.011	0.010	–1.53	0.22	–1.00
F7	0.64	2.24	1.43	3.6	0.06	0.046	0.025	0.018	0.011	0.008	–0.91	0.41	–0.37
F8	0.76	2.05	1.56	5.8	0.08	0.048	0.041	–0.020	0.004	–0.007	–1.49	0.63	–0.97
F9	0.77	2.14	1.65	3.3	0.02	0.048	0.020	–0.013	0.013	0.013	–1.63	0.61	–1.68
F10	0.72	1.96	1.41	–6.5	0.07	0.048	0.029	–0.015	0.008	0.007	–1.68	0.33	–1.97
F11	0.77	1.93	1.49	2.0	0.09	0.040	0.024	–0.012	0.007	0.005	–1.64	0.41	–1.85
F12	0.70	1.80	1.25	2.7	0.12	0.035	0.029	–0.013	0.002	0.004	–1.59	0.87	–1.80
F13	0.64	1.97	1.26	3.9	0.11	0.064	0.059	–0.025	0.011	–0.008	–1.43	1.20	–2.06
Mean	**0.75**	**1.88**	**1.41**	**6.03**	**0.079**	**0.066**	**0.043**	**–0.017**	**0.003**	**–0.002**	**–1.26**	**0.70**	**–1.51**
Max.	**0.85**	**2.24**	**1.65**	**14.8**	**0.169**	**0.11**	**0.08**	**0.028**	**0.015**	**0.013**	**0.30**	**1.44**	**1.08**
Min.	**0.57**	**1.64**	**1.00**	**–6.5**	**0.024**	**0.04**	**0.02**	**–0.04**	**–0.015**	**–0.017**	**–1.68**	**–0.22**	**–2.23**
Standard deviation	**0.07**	**0.13**	**0.15**	**4.71**	**0.03**	**0.02**	**0.01**	**0.02**	**0.01**	**0.01**	**0.45**	**0.54**	**0.87**

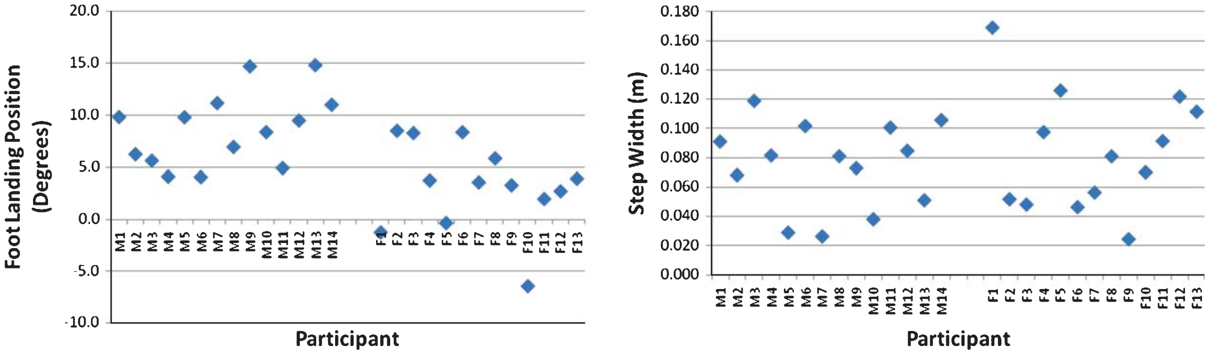

Fig. 6. Distribution of measured values for foot landing position and step width respectively.

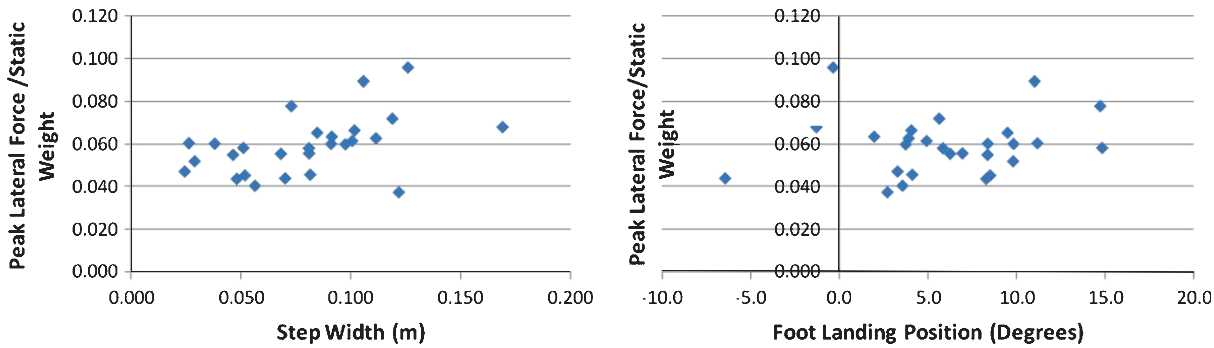

Fig. 7. Relationship of peak lateral force/static weight to step width and foot landing position respectively.

suggested as influencing the lateral force exerted by pedestrians. However, the recorded values for both of these parameters all fell within normal ranges and there was not enough evidence from these trials to support the theories that they impact on lateral loading. Figure 7 shows the relationship between the ratio of peak lateral force to static weight and the step width and foot landing position respectively. Further analysis of the impact of these parameters therefore is not considered in this paper.

4.2. Individual GRF results

Ground reaction force (GRF) results in the medio-lateral plane were recorded for each crossing by each individual test participant. A total of over 100 such traces were recorded and displayed considerable intra-participant similarity, so for indicative purposes only one such result will be discussed herein. Figure 8(a) shows the four recorded GRF plots for the male participant, M6. It can be seen that the typical shape of these plots is an initial medial force exerted on the ground as the foot makes contact, followed by a lateral force characterized by two main peaks, as the foot moves to propel the pedestrian forward. It is observed that the magnitude of the peak medial force is approximately equal to the magnitude of the peak lateral reaction force. The uniformity of the traces should also be noted. The traces shown in Fig. 8(a) include two from the left foot and two from the right foot. The two from the right foot have been inverted to simplify comparison between all four traces. This also illustrates the similarity between the reaction forces from the right and left foot respectively, as would be expected from a healthy test subject. The mean peak value of the lateral force from a single footfall normalized by the pedestrian's static weight is 0.059 (or 5.9%), which is slightly above the range of 4–5% as reported in Section 1.

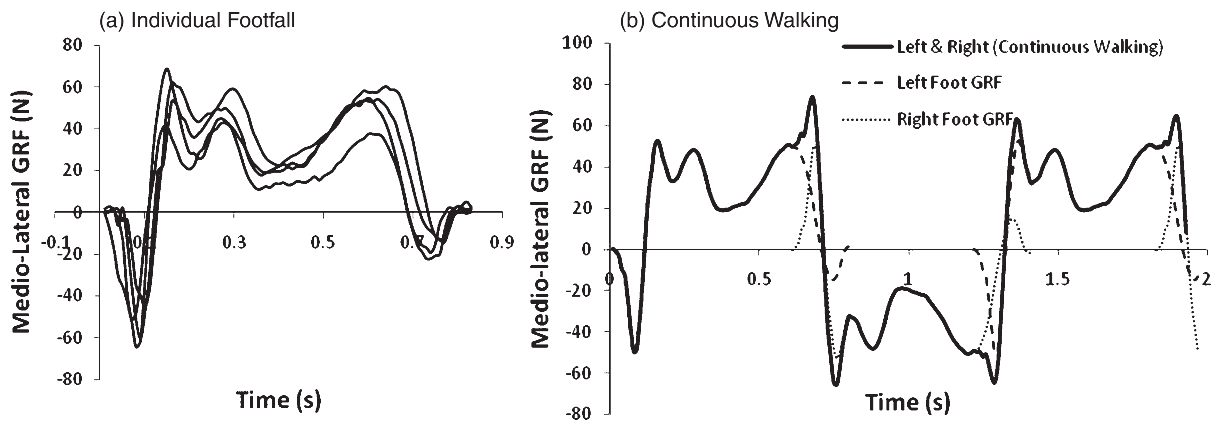

Fig. 8. Medio-lateral ground reaction force traces measured from participant M6.

4.3. Continuous medio-lateral force estimation

For each participant, an average medio-lateral footfall trace was computed based on the number of full datasets for the individual. The continuous lateral reaction force trace for consecutive footsteps was estimated by consecutively summing the appropriate value of these average force traces to represent the left and right feet respectively. It is assumed in this approach that the force traces from the left and right feet respectively are identical in magnitude and inverted in direction. The time separation between footsteps was determined from the inverse of the mean pacing frequency for each test participant. Figure 8(b) shows a typical continuous lateral ground reaction force plot. Note that there is an overlap between the individual footfall traces and this represents the double-stance phase of human walking i.e. the time when both feet are in contact with the walking surface. This continuous lateral force trace will be referred to as the measured trace.

4.4. Lateral force simulation

Two approaches have been adopted to simulating the recorded lateral force trace in this paper. Firstly, a simplistic, single harmonic sine function was developed. Secondly, a multiple-harmonic function was developed and calibrated against the recorded data.

4.4.1. Single harmonic sine function

Other authors have proposed lateral load models of the form given in equation 1, whereby the applied load varies sinusoidally with respect to time. The frequency of force application is half the pacing frequency considered. For each of the test subjects, a similar function was derived and the variable L_f, the dynamic load factor, was optimized through matching the peak magnitudes of the ‘measured’ and simulated responses. Table 2 contains the optimum values for these load factors. It can be seen that the dimensionless dynamic load factor ranges from 0.035 (or 3.5%) to 0.106 (or 10.6%), with a mean for the entire test population of 0.066 (6.6%). This is higher than the mean value if only the GRF trace from one foot is considered, but reflects the effect of the superposition of medial forces from one step on the lateral force from the previous footfall as a result of the double stance phase of walking. This value is also higher than previously published values of 3% [24] and 4% [3]. It is also worth noting that the mean value for the male responses was 0.077, while for the female results, the mean was only 0.055. This could potentially indicate a significant gender difference in the generation of lateral forces and is worthy of further investigation.

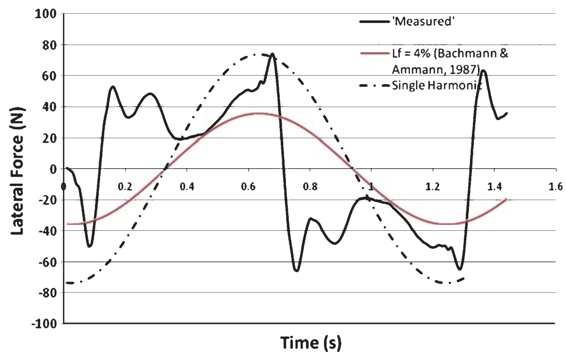

Fig. 9. Lateral force recorded and simulated using single harmonic load function for M12.

Figure 9 shows the ‘measured’ and simulated lateral forces for male test subject, M12, using a single harmonic force function. Also shown is the single harmonic force function with a value of 4% for the dynamic load factor as proposed by Bachmann and Ammann [3]. Single harmonic force functions, while capturing the peak values of the lateral force and obviously the frequency of force application, do not reflect the true nature of this loading regime. The potential for employing a multiple harmonic force function was thus investigated.

4.4.2. Multiple harmonic sine function

A multiple harmonic force function was developed to simulate the recorded lateral force trace. In considering this function, the contributions from even number multiples of the fundamental frequency were omitted as their contribution was deemed negligible. This is consistent with results reported Erlicher et al. [12] and differs from the work of Bachmann and Ammann [3] who propose DLF values for all harmonics. In this way, the function differs from that in equation 2. Moreover, further analysis of the individual dynamic load factors is carried out in this work based on the responses from over 100 walking trials involving 27 participants. This offers a more statistically robust offering than that of Erlicher et al. [12]. Thus the multiple harmonic function consisted of a fundamental frequency component which was equal to half the mean pacing frequency for the respective pedestrian and the third, fifth and seventh harmonic of this frequency. Higher order harmonics were shown to have negligible impact. This function is represented by equation 3.

$$
\begin{aligned}
F(t) &= \sum_{i=1}^{7} L_{fi}\, G \sin(i\pi f_s t + \phi_i) \quad i = 1, 3, 5, 7 \\
&= L_{f1}\, G \sin(i\pi f_s t) + L_{f3}\, G \sin(3\pi f_s t + \phi_3) \\
&\quad + L_{f5}\, G \sin(5\pi f_s t + \phi_5) + L_{f7}\, G \sin(7\pi f_s t + \phi_7)
\end{aligned} \tag{3}
$$

The values of the respective dynamic load factors and phase angles were optimized by minimizing the sum of the square of the error between the measured and simulated responses, while using the Fourier decomposition of the response signal. This optimization was performed in Microsoft Excel, using the Solver function with a tolerance of 5% and a convergence value of 0.0001. Table 2 shows the final values for each of these parameters, with mean values of 0.043 and 0.017 for the 1st and 3rd harmonic respectively. The maximum values for these two parameters were 0.080 and 0.028 respectively and these maxima could serve as conservative input values for practitioners attempting to use this model. As before, the values for the male subjects are higher than for the females. These results indicate a significant contribution to the lateral force function

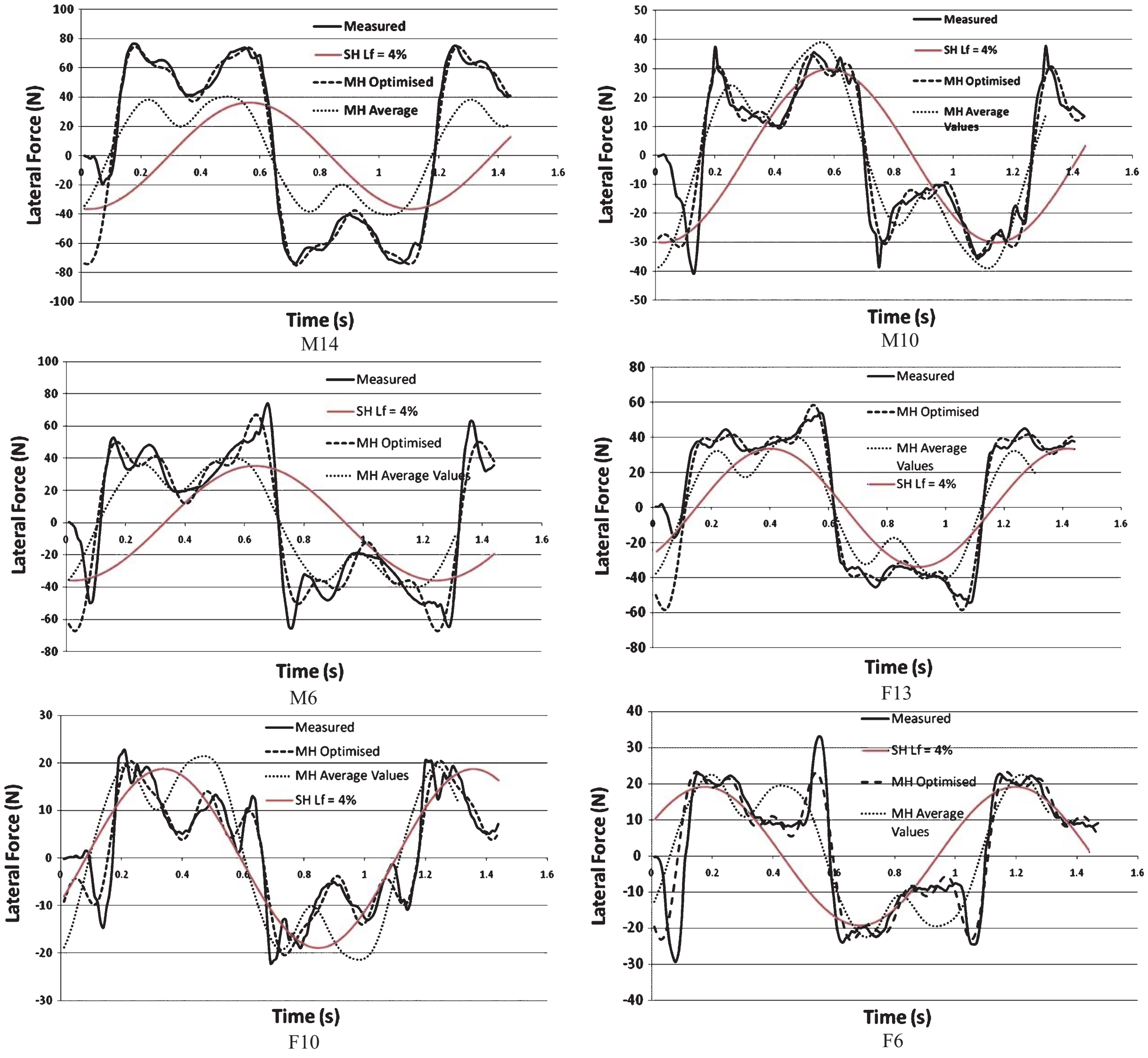

Fig. 10. Selected recorded and simulated lateral force functions.

at the third harmonic of the fundamental frequency or $1.5f_s$. This is in agreement with the work of Nakamura et al. [21], who reported a significant contribution at the third harmonic based on frequency domain analysis of the lateral force.

The values for the 5th and 7th harmonics are close to zero, so future simulations may only focus on the first and third harmonics of the fundamental frequency. Figure 10 shows the measured and simulated lateral force functions for M14, M10, M6 and F13, F10, F6 respectively for illustrative purposes. These show excellent agreement between the measured responses and those simulated using the optimum values for dynamic load factors and phase angles.

Also shown are the simulated responses using a single harmonic (SH) function with a dynamic load factor of 4%, together with a multiple harmonic (MH) function employing the mean values for the dynamic load factors and phase angles as contained in Table 2. This MH function with mean input values offers a reasonable attempt to simulate each of the responses and may have potential as an input function into attempts to model lateral loading from crowds of pedestrians.

5. Conclusions

Lateral excitation of flexible structures such as footbridges by crossing pedestrians is the subject of extensive research at present. Despite this, there is still relatively little published data on the exact nature of the lateral force function directly generated by individual pedestrians. This paper reports on data obtained from over 100 walking trials by a healthy adult test population of 14 males and 13 females walking at self-selected nominal 'normal' velocity along an 11 m walkway.

The mean values for pacing frequency, foot landing position and step width measured during the trials were 1.88 Hz, 6.03° and 0.079 m respectively. Analysis of any relationship between the lateral forces and step width or foot landing position is not contained in this work.

Mean values for the dynamic load factor associated with a single harmonic sine function range from 0.055 for the female participants to 0.077 for the males, with an overall mean of 0.066 (or 6.6%). This function can simulate the peak magnitudes of the lateral forces but fails to capture the obvious higher order frequency components.

A multiple harmonic (MH) function was thus developed considering the first seven harmonics of the fundamental frequency of force application, which is half the pacing frequency, f_s. The contribution from even numbered harmonics was seen to be negligible thus these harmonics were omitted from the final function. Optimum values for the dynamic load factors and phase angles associated with each harmonic were determined through minimizing the error between the measured and simulated responses. This yielded values of 0.043, −0.017, 0.003 and 0.002 for L_{F1}, L_{F3}, L_{F5} and L_{F7} respectively, indicating a relatively significant contribution from the third harmonic of the fundamental frequency, which corresponds to a frequency of $1.5f_s$.

Simulations were then carried out using the mean values for dynamic load factors and phase angles determined from all of the tests and these showed reasonably good agreement with all of the measured functions.

For the load model considered, only the first and third harmonics were significant. Conservatively and in the absence of data from a greater sample population, practitioners and other researchers could use the derived maximum values for the first and third dynamic load factors of 0.08 and 0.028 respectively. Using these maxima naturally overestimates the lateral forcing function in most cases.

The influence of gait parameters and other parameters such as walking velocity were not examined here but are the subject of current work by the authors.

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