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Full Length Article

Keep your head down: Maintaining gait stability in challenging conditions

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A B S T R A C T

Background: Peripheral vision often deteriorates with age, disrupting our ability to maintain normal locomotion. Laboratory based studies have shown that lower visual field loss, in particular, is associated with changes in gaze and gait behaviour whilst walking and this, in turn, increases the risk of falling in the elderly. Separately, gaze and gait behaviours change and fall risk increases when walking over complex surfaces. It seems probable, but has not yet been established, that these challenges to stability interact.
Research question: How does loss of the lower visual field affect gaze and gait behaviour whilst walking on a variety of complex surfaces outside of the laboratory? Specifically, is there a synergistic interaction between the effects on behaviour of blocking the lower visual field and increased surface complexity?
Methods: We compared how full vision versus simulated lower visual field loss affected a diverse range of behavioural measures (head pitch angle, eye angle, muscle coactivation, gait speed and walking smoothness as measured by harmonic ratios) in young participants. Participants walked over a range of surfaces of different complexity, including pavements, grass, steps and pebbles.
Results: In both full vision and blocked lower visual field conditions, surface complexity influenced gaze and gait behaviour. For example, more complex surfaces were shown to be associated with lowered head pitch angles, increased leg muscle coactivation, reduced gait speed and decreased walking smoothness. Relative to full vision, blocking the lower visual field caused a lowering of head pitch, especially for more complex surfaces. However, crucially, muscle coactivation, gait speed and walking smoothness did not show a significant change between full vision and blocked lower visual field conditions. Finally, head pitch angle, muscle coactivation, gait speed and walking smoothness were all correlated highly with each other.
Significance: Our study showed that blocking the lower visual field did not significantly change muscle coactivation, gait speed or walking smoothness. This suggests that young people cope well when walking with a blocked lower visual field, making minimal behavioural changes. Surface complexity had a greater effect on gaze and gait behaviour than blocking the lower visual field. Finally, head pitch angle was the only measure that showed a significant synergistic interaction between surface complexity and blocking the lower visual field. Together our results indicate that, first, a range of changes occur across the body when people walk over more complex surfaces and, second, that a relatively simple behavioural change (to gaze) suffices to maintain normal gait when the lower visual field is blocked, even in more challenging environments. Future research should assess whether young people cope as effectively when several impairments are simulated, representative of the comorbidities found with age.

1. Introduction

Maintenance of our stability when walking depends heavily on processing visual information from the environment. Visual information is particularly important for stability when the environment is more complex, leading to people modifying locomotion in real-time for safe navigation (Marigold & Patla, 2007; Matthis, Barton, & Fajen, 2017; Matthis, Yates, & Hayhoe, 2018; Patla & Greig,

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2006). We can assess how different environments (and other factors) affect stability by measuring gaze and gait behaviour. Although no stability measure has been accepted as a gold standard (as reviewed in Bruijn, Meijer, Beek, & Van Dieën, 2013), assessing a range of gaze and gait behavioural changes allows the researcher to build up a portfolio of how the body adapts to a given manipulation, such as increasing surface complexity, providing converging evidence about its effect on stability when walking. Such changes may include "dangerous" behaviours that increase fall risk as well as "cautious" behaviours that occur to try to reduce fall risk. For example, when the perceived risk of a fall increases, people have been found to adopt a more cautious gait, characterised by a slower gait with shorter and wider steps (Pirker & Katzenschlager, 2017). Unfortunately, some of these cautious behavioural changes that are intended to reduce fall risk may be dangerous in that they can lead to increased fall risk. For example, increased leg muscle coactivation (simultaneous contraction of an agonist and antagonist muscle) helps to stabilise the leg when walking (Thompson, Plummer, & Franz, 2018). However, a stiff leg is less flexible and therefore has a reduced range of motion which, in itself, is a known risk factor for falls (Chiacchiero, Dresely, Silva, DeLosReyes, & Vorik, 2010; Reddy & Alahmari, 2016). Regardless of whether the behavioural response is intentional, these behavioural changes indicate deviations from normal, stable gait, and thus can help to identify the factors that influence stability and perceived fall risk in a given situation.

When people walk over more complex surfaces a number of different behavioural changes occur. Here we define complex surfaces as any non-smooth surface including slope changes, uneven surfaces, stairs and inconsistently spaced foot targets (Thomas, Gardiner, Crompton, & Lawson, 2020a; Thomas, Gardiner, Crompton, & Lawson, 2020b). Compared to smooth, level walking, complex surfaces are associated with reduced step length, increased step width variability, increased leg muscle coactivation and reduced gait speed (Marigold & Patla, 2008a; Menant, Steele, Menz, Munro, & Lord, 2009; Thomas et al., 2020a; Voloshina, Kuo, Daley, & Ferris, 2013). Walking on stairs compared to smooth surfaces is associated with increased anteroposterior sway at the lower back and increased step variability (Wang et al., 2014; Wang et al., 2017). Gaze (combined eye and head movements) alters when walking over more complex surfaces, with increased fixations and gaze directed closer toward the person's feet (t Hart & Einhauser, 2012; Marigold & Patla, 2007; Matthis et al., 2018; Thomas et al., 2020a). We have developed a multimethod approach to measure surface complexity in order to try to characterise surfaces in terms of behaviour indicative of stability for walking (Thomas et al., 2020b). We assessed how physical and perceptual measures of surface complexity across a wide range of surfaces influenced gait and gaze. Using these measures we found that head pitch lowered, muscle coactivation increased and walking symmetry reduced when walking over more complex surfaces.

In young healthy individuals, environmental information can be obtained from peripheral vision (vision outside the centre of gaze fixation). For example, young, healthy individuals can walk over unexpected objects even when they are fixating well above the ground plane such that the obstacles are only visible in the periphery of their lower visual field (Franchak & Adolph, 2010; Marigold & Patla, 2007). However, as people age, both the rate of comorbidities associated with vision loss increase and healthy peripheral vision is known to deteriorate (Beurskens & Bock, 2012; Collins, Brown, & Bowman, 1989; Crassini, Brown, & Bowman, 1988). Given the loss of peripheral vision, visual perception may be disrupted, thus increasing the challenges to the elderly in maintaining stable locomotion, especially over more complex surfaces. For example, lower visual field loss is a symptom of glaucoma, an eye condition that is particularly common in the elderly. Lower visual field loss due to glaucoma is associated with an increased rate of falls (Black, Wood, & Lovie-Kitchin, 2011) and people with glaucoma exhibit increased step to step variability in step length, make more erroneous steps, have a slower gait, and fixate closer to their feet (Friedman, Freeman, Munoz, Jampel, & West, 2007; Lajoie, Miller, Strath, Neima, & Marigold, 2018; Mihailovic et al., 2017; Miller, Lajoie, Strath, Neima, & Marigold, 2018). Another eye condition associated with lower visual field loss, retinitis pigmentosa, results in individuals fixating at the ground for longer when level walking compared to those with normal vision (Timmis et al., 2017). Studies conducted outside have shown that those with peripheral visual field loss from glaucoma or retinitis pigmentosa make more errors when judging gaps in traffic at pedestrianised crossings compared to those with normal vision (Cheong, Geruschat, & Congdon, 2008). In contrast, eye diseases associated with central visual field loss (e.g. macular degeneration) appear to have less effect on gait, with only a reduction in speed shown when negotiating a curb compared to those with normal vision (Alexander et al., 2014).

A complicating factor in interpreting these results is that many of these eye diseases are more prevalent with age, and thus study participants are typically older and are likely to have other perceptual, cognitive and musculoskeletal deficits. One alternative, to assessing peripheral visual field loss is to simulate its loss in young individuals who do not suffer from these confounding factors. For young individuals, simulation of lower visual field loss by wearing goggles with the lower area blocked, has been shown to lead to the adoption of a more cautious, slower gait and reduced foot placement accuracy (Graci, Elliott, & Buckley, 2010; Marigold & Patla, 2008b; Rietdyk & Drifmeyer, 2009), all of which are suggestive of a less stable gait. This use of goggles to simulate lower visual field loss, due to eye disease and ageing, has the advantage of being a relatively easy manipulation, however, goggles do not have an identical effect to the eye diseases experienced commonly in the elderly. This is due to the fact that the goggles move with the head, not the eyes, and thus they block a variable amount of the lower visual field. In contrast lower visual field loss due to eye disease blocks information from a constant area of the visual field. Nevertheless, importantly, lower visual field loss from either wearing goggles or eye diseases will typically lead to less information being available from the lower portions of the scene and, in particular, the area around the feet, unless compensatory movements are made, including tilting the head downwards.

In summary, there is undoubtedly a lack of understanding about how peripheral visual field loss affects behaviour whilst walking. Loss of the lower visual field, in particular, appears to increase fall risk whilst walking (Black et al., 2011; Graci et al., 2010; Marigold & Patla, 2008b; Rietdyk & Drifmeyer, 2009). However, it remains unclear how the lower visual field influences walking over surfaces representative of those typically encountered outside of the gait laboratory. Here we investigate how combining a simulated lower visual field loss with walking over more complex surfaces changes gaze and gait behaviour. This is critical given that populations who are most vulnerable to falling, such as the elderly, often suffer from multiple challenges simultaneously. These challenges are both intrinsic, due to deteriorating perception, cognition or musculoskeletal function, and extrinsic, due to testing everyday situations such as uneven or slippery surfaces, poor lighting and crowded, fast-changing environments. In order to isolate the effects of visual field loss and surface complexity on walking behaviour, in the present study, we tested young, healthy individuals who were not suffering from comorbidities. Young people walked over a wide variety of outside surfaces with full vision and whilst wearing goggles which blocked their lower visual field. The surfaces included those that have previously been categorised as smooth, irregular and stairs (Thomas et al., 2020b). We measured gaze behaviour (head pitch angle, eye angle) and gait behaviour (muscle coactivation, gait speed and walking smoothness as measured by harmonic ratios). We determined how these measures responded to a perceptual measure of surface complexity (see Thomas et al., 2020b). In combination, these gaze and gait measures allowed us to investigate how people respond when challenged by walking over diverse surfaces when the lower visual field is blocked.

2. Methodology

2.1. Participants

Twenty healthy adults (13 male, mean ± SD; age = 26.42 ± 4.27 years; height = 176.1 ± 9.33 cm) were recruited for the study. No participant had any known impairment or injury which might affect their gait or vision. All of the inertia data that was used to identify gait behaviours did not record properly for one participant, so that gait data was only analysed in 19 participants. Furthermore, the inertia data that was collected at the lower back (used to calculate harmonic ratios to provide a measure of walking smoothness) did not record properly for two further participants, and for one of those participants the inertia data collected at the ankle (used to calculate gait speed) did not record properly, thus only 17 participants and 18 participants respectively are included in these particular analyses.

2.2. Data collection

Ethical approval was granted for the study in November 2017 by the University of Liverpool's Ethics Committee (REF: 2672). Five behavioural measures were assessed: head pitch angle, eye angle, muscle coactivation, gait speed and walking smoothness as measured by harmonic ratios. We also measured the duration and number of eye fixations to provide a check of whether eye movements were influenced by wearing the goggles. Eye angle, calculated from vertical pupil movements, and the duration and number of eye fixations, were recorded using a Pupil Labs eye-tracker (Kassner, Patera, & Bulling, 2014). Head pitch, muscle coactivation, gait speed and harmonic ratios were recorded from six Delsys TRIGNO™ Inertia Measurement Unit (IMU) sensors (Boston, MA, USA) placed on participants. Four of these sensors collected inertial data (148 Hz) at the head, lower back and superior to both ankles, whilst two sensors collected surface electromyography (sEMG) data (1111 Hz) from the *Tibialis Anterior* and medial head of the *Gastrocnemius* muscle of the left leg. Further details of data collection are given in the supplementary material (SM1). The lower visual field was blocked using basketball goggles, following the same technique as Rietdyk and Drifmeyer (2009). Fig. 1A

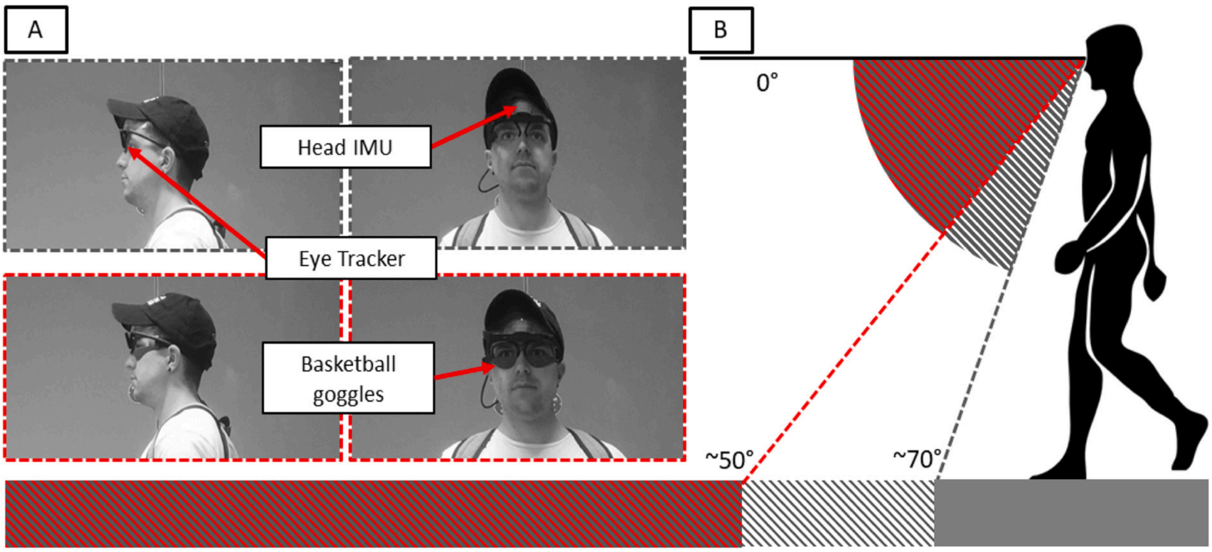

Fig. 1. (A) Images showing the experimental set-up at the head for the full vision (top) and blocked lower visual field conditions (bottom). Participants wore an eye tracker (used to record eye movements), a head IMU sensor (used to calculate head pitch), basketball goggles (used to block the lower visual field) and a baseball cap (used to shade the eye-tracker during outdoor testing). (B) A diagram showing the approximate ranges of negative eye angles from which information about the upcoming surface could be extracted when the head was level. For full vision conditions, negative eye angle ranged from 0° to −70°, (regions striped grey and red), whilst for blocked lower visual field conditions, negative eye angles ranged from 0° to −50° (regions striped red only), see supplementary materials (SM2) for details. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

shows an example of the experimental set-up of the head for both full vision and blocked lower visual field conditions. Fig. 1B shows the approximate extent to which the goggles blocked the participants' view of lower areas of the scene. Compared to full vision (no goggles), the goggles blocked approximately the lowest 20° of vision when the head was level (see supplementary material SM2 for details).

2.3. Protocol

Participants walked over all of the surfaces with full vision and with a blocked lower visual field. For each of the two vision conditions, participants walked over 14 different surfaces at a self-selected speed. Assignment to the initial vision condition was alternated across participants.[1] As such participants completed either the blocked lower visual field condition first (and, here, completed surfaces in order from A to N), whilst the remaining participants completed the blocked lower visual field condition second (and thus completed surfaces in order from N to A), (see supplementary materials SM3). The surfaces were located across the University of Liverpool campus and are shown in Fig. 2. The total duration of the study (including debriefing, marker placements, calibrations and the trials themselves) for each participant was approximately 150 min of which over 80 min was data collection.

For each surface, participants were instructed to look straight ahead whilst standing still in front of the surface for three seconds, then they walked at a comfortable speed across the surface. At the end of the surface, participants again looked straight ahead whilst standing still for a further three seconds. The periods spent looking straight ahead at the start and end of the trial were used to remove drift from the gyroscopic data recorded at the head IMU, vertical gyroscopic data used to calculate head pitch angle. Other than at the start and end of each surface, participants were told that they should move their eyes and head as normal whilst walking.

We tried to ensure that all participant completed the study under similar conditions. Specifically, all participants were tested in the summer, when the University campus was relatively quiet (mid-morning or mid-afternoon), and on dry days. Nevertheless, compared to laboratory based testing, testing conditions were more variable such that our findings should be relatively robust and relevant to walking outdoors.

Surface complexity for each surface was measured using perceptual ratings of roughness and two ratings of perceived stability as detailed in Thomas et al. (2020b). Participants[2] rated surfaces on a Likert scale (Likert, 1932) between 1 (smooth / stable) and 10 (rough / unstable) with participants rating surfaces from vision alone, for perceived roughness and perceived stability, and then participants re-rated perceived stability after having walked on the surface. Surface complexity was taken as the average of the three ratings, given that the ratings were highly correlated (r from 0.94–0.98). Fig. 2 shows surfaces ranked from the easiest (S1) to most complex (S14). Perceptual ratings were used as a measure of surface complexity, rather than grouping surfaces together based on similar physical features (smooth, irregular, stairs etc.) due to the inconsistent terminology in the literature using such descriptors (Marigold & Patla, 2007; Matthis & Fajen, 2014; Merryweather, Yoo, & Bloswick, 2011; Patla & Vickers, 2003; Thies, Richardson, & Ashton-Miller, 2005), see Thomas et al. (2020b) for further discussion. In the present approach, surface complexity is based solely on the participants' evaluations and not on our own.

2.4. Analysis

For each surface, mean head pitch angle, eye angle, muscle coactivation, gait speed and walking smoothness as measured by harmonic ratios were calculated.[3] Head pitch angles were calculated from gyroscopic data from the head IMU with 0° defined as the average position during the three seconds that the participant was static at the start and end of each surface trial. Eye angles of 0° were defined from participant's fixating at a target set at their eye height during calibration of the eye tracker. Deviations from 0° eye angle due to vertical eye movements were converted into angles with eye movements down taken as negative angles. Only eye angles within the normal range expected were analysed (Lee, Kim, Shin, Hwang, & Lim, 2019), for further details see supplementary material (SM2). In addition eye fixation duration and number of eye fixations were recorded, with fixations defined as stabilised eye movement for at least 100 milliseconds following that of previous research (Marigold & Patla, 2007; Patla & Vickers, 1997, 2003). The average number of eye fixations per metre walked was used to avoid differences caused by the surfaces being different lengths. We also calculated mean relative frequency distributions of head pitch and eye angles for each surface under full vision and blocked lower visual field conditions. Frequencies of head pitch and eye angles were recorded in bins of 5° for each surface. This method follows that of Foulsham, Walker, and Kingstone (2011); Thomas et al. (2020a). Muscle coactivation was calculated following Winter

[1] As the attachment and removal of sensors and initial eye calibration had to be completed in the gait laboratory, for practical reasons of locations and sensor battery life, all participants started by completing surfaces from A through to N and then they completed surfaces N though to A.

[2] Full details of the perceptual rating study are provided in Thomas et al. (2020b). In brief, 32 participants (14 male, mean ± SD; age = 22.2 ± 5.0 years; height = 172.6 ± 8.5 cm), completed the perception rating study. Twelve of these participants had been participants in the present study (10 male, age = 27.3 ± 4.3 years; height = 178.0 ± 6.9 cm). There were no significant differences between the mean responses for the participants who completed both studies and the remaining 20 participants, ($F(1,32) = 0.22$, $\eta_p^2 = 0.01$, $p = 0.643$) so responses were pooled together.

[3] One limitation of the study was that we were only able to collect data for participants walking once over a given surface for each of the two visual conditions. This was due to the time taken to complete the study and the limited battery life for the IMU sensors used in the study. However, for each condition and for all surfaces we collected eye, inertia and EMG data at high frequencies (between 30 Hz and 1111 Hz) for our five behavioural measures, for each condition and for all surfaces. We collected over 80 min of data per participant and mean values per trial were based on averages over hundreds of values.

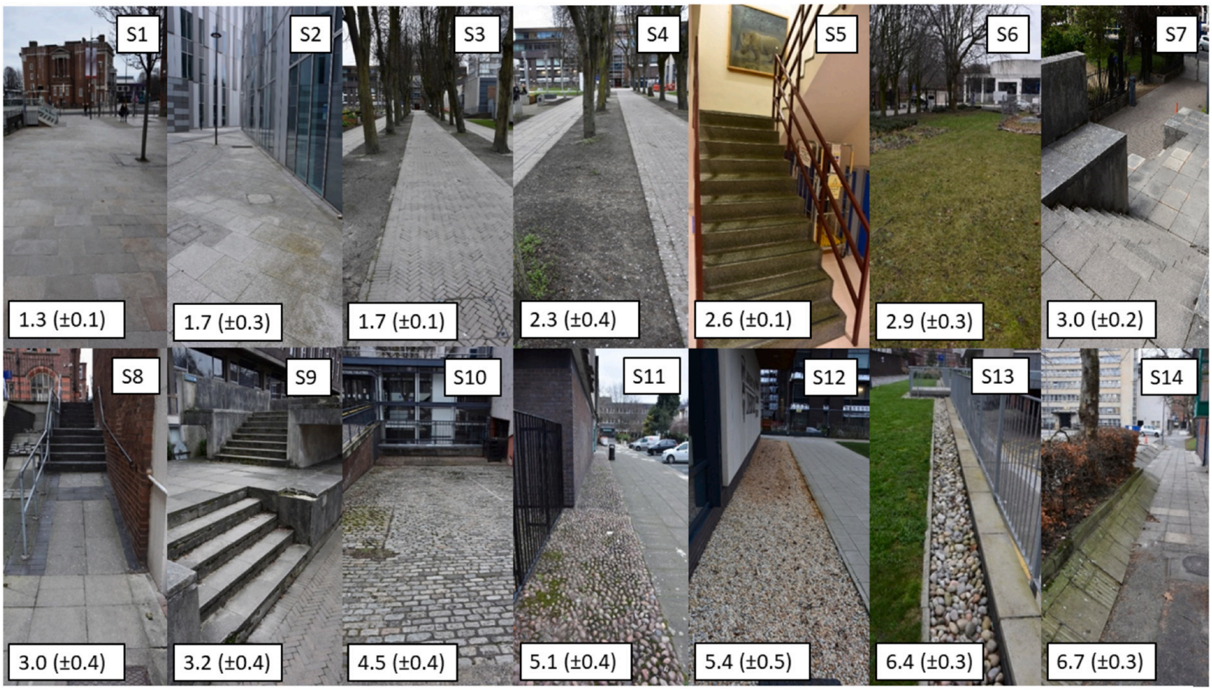

Fig. 2. Images showing the 14 surfaces used in the study. Surfaces were ranked based on participants' average perceptual ratings of surface complexity from S1 (smoothest / most stable) to S14 (most complex / hardest to walk over). Average (± SE) perceptual ratings are shown for each surface. See supplementary materials (Table SM3) for more information.

(2005) for the *Tibialis anterior* and medial head of the *gastrocnemius* muscles across each gait cycle. Gait speed was calculated from gyroscopic data at the ankle following the method from Li, Young, Naing, and Donelan (2010) and known surface lengths, see supplementary material (SM3). Walking smoothness was measured in terms of harmonic ratios which were calculated from anteroposterior accelerometery data from the IMU placed at the lower back. A higher ratio was interpreted as a more symmetrical, smoother gait following Bellanca, Lowry, VanSwearingen, Brach, and Redfern (2013).

In order to show how behavioural metrics related to surface complexity, regression analyses were conducted on the participants' mean head pitch angle, eye angle, muscle coactivation, gait speed and walking smoothness. This was done separately for full vision and blocked lower visual field conditions with the independent variable of surface complexity perceptual rating. To compare behavioural changes between the two vision conditions, we conducted *t*-tests between the regressions' intercepts and slopes. Significant differences were taken as $p < 0.05$. Finally, we conducted Pearson's correlations on the mean z-scores of the different measures for each surface. The z-scores for muscle coactivation measures were multiplied by -1 so that, for all five measures, higher z-scores were associated with more stable walking. Large correlations ($r > 0.5$, as determined by Cohen (2013)) are shown in bold for each correlation table. A conservative alpha level of 0.001 was used for correlations, calculated using the Bonferroni correction.

3. Results

The behavioural data first reported in Thomas et al. (2020b) is presented here again as data for full vision conditions. In our previous work this data was used in conjunction with physical and perceptual metrics to analyse different aspects of surface complexity. In the present paper the full vision data was used in combination with previously unpublished data, namely that obtained in the blocked lower visual field condition, to investigate the effect on gait and gaze behaviour of both the availability of visual information and different walking surfaces. Initial analyses revealed that gaze and gait behaviour for the four surfaces with stairs (S5, S7, S8, and S9) differed markedly from that of other surfaces. Despite stairs being typical surfaces found in everyday environments, stairs differ in terms of biomechanics and muscle activation relative to walking over other surface (Cromwell & Wellmon, 2001; Wang et al., 2017; Zietz & Hollands, 2009). Our results presented here are consistent with that of the previous literature. Due to the distinct patterns shown for stairs, the analysis presented here excluded those four surfaces. We return to address the issue of stairs in the general discussion. To aid comparisons, surfaces with stairs are still plotted on the figures and, for the interested reader, full analyses including stairs are given in the supplementary material (SM4).

3.1. Head pitch angle

Mean head pitch angles for full vision and blocked lower visual field conditions are shown in Fig. 3A. A linear regression revealed

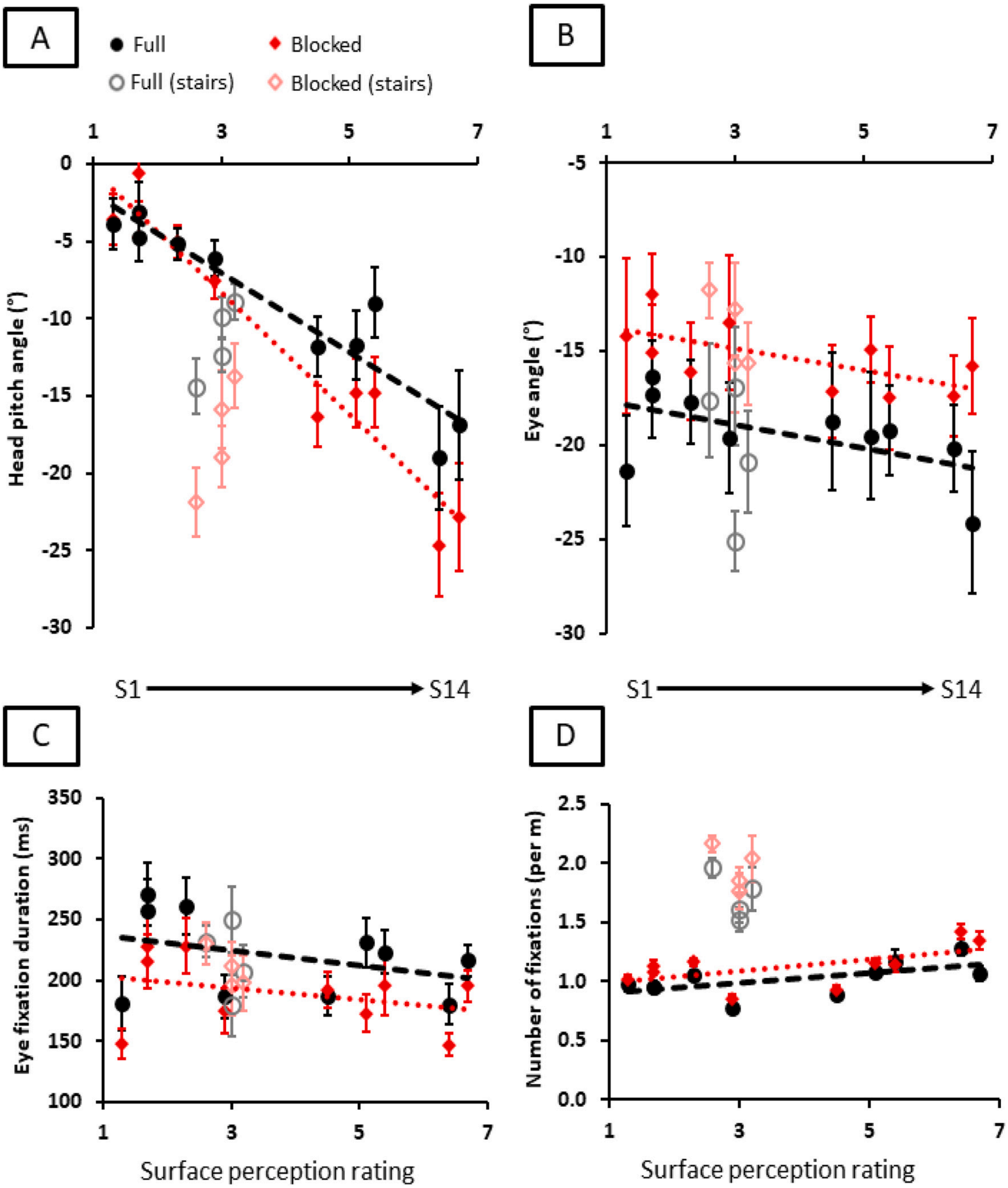

Fig. 3. Mean (± SE) (A) head pitch angles, (B) eye angles, (C) eye fixation duration, and (D) number of eye fixations per metre walked for surfaces S1 – S14 for full vision (black circles) and blocked lower visual field (red diamonds) conditions. Surfaces with stairs are represented separately (grey open circles and light red open diamonds for full vision and blocked lower visual field respectively). Dotted lines represent the regression lines (when excluding stairs) for full vision and blocked lower visual field conditions. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

a significant relation between head pitch and surface complexity for both full vision and blocked lower visual field conditions ($R^2 = 0.881$; $F,(1,8) = 59.43$, $p < 0.001$ and $R^2 = 0.939$; $F,(1,8) = 122.16$, $p < 0.001$) respectively. A *t*-test between the regression intercepts was not significant, *($t(19) = 1.43$, $p = 0.173$),* however, there was a significant difference between regression slopes *$t(19) = -2.88$, $p = 0.010$)*. Head pitch angle for blocked lower visual field conditions showed a greater decrease than that for full vision conditions as surface complexity increased. On the simplest surfaces, head pitch for both conditions was around $-5°$. However, on the most complex surfaces, head pitch for blocked lower visual field conditions ($-22.8°$) was around 6° lower than for full vision (−16.8°). Thus with a blocked lower visual field participants lowered their head more when surfaces were more complex, compared to with full vision.

3.2. Eye angle

Mean eye angle for full vision and blocked lower visual field conditions are shown in Fig. 3B. A linear regression showed no significant relation between eye angle and surface complexity for full vision ($R^2 = 0.33$; $F,(1,8) = 3.91$, $p = 0.083$), however, the equivalent regression for blocked lower visual field conditions was significant ($R^2 = 0.43$; $F,(1,8) = 6.05$, $p = 0.039$). A t-test between the regression intercepts was also significant, ($t(19) = 2.34$, $p = 0.033$), but there was no significant difference between regression slopes ($t(19) = 0.11$, $p = 0.916$. On average eye angle for full vision (−19.4°) was 4° lower than with blocked lower visual field conditions (−15.4°). However, importantly, this behavioural change occurred irrespective of surface complexity, see Fig. 3B.

3.3. Duration and number of eye fixations

We completed additional analysis to assess the duration and number of eye fixations to check whether these were influenced by blocking the lower visual field. Mean eye fixation duration and the number of eye fixations per metre walked for full vision and blocked lower visual field conditions are shown in Figs. 3C & 3D. A linear regression showed no significant relation between the duration of eye fixations and surface complexity for either full vision or blocked lower visual field conditions ($R^2 = 0.13$; $F,(1,8) = 1.24$, $p = 0.299$ & $R^2 = 0.11$; $F,(1,8) = 0.97$, $p = 0.354$ respectively). There were also no significant differences between the regression intercepts ($t(19) = -1.14$, $p = 0.273$) or between regression slopes ($t(19) = 0.21$, $p = 0.833$. Similarly, a linear regression showed no significant relation between the number of eye fixations and surface complexity for either full vision or blocked lower visual field conditions ($R^2 = 0.37$; $F,(1,8) = 4.69$, $p = 0.062$ & $R^2 = 0.35$; $F,(1,8) = 4.346$, $p = 0.071$ respectively). There was also no significant differences between the regression intercepts ($t(19) = 0.58$, $p = 0.569$) or between regression slopes ($t(19) = 0.20$, $p = 0.846$). Thus increasing surface complexity did not change eye fixation duration or number for either full vision or blocked lower visual field conditions.

Frequencies of eye and head pitch angles were recorded in bins of 5° for each surface following the method of Foulsham et al. (2011); Thomas et al. (2020a). The mean frequency distribution for these 5° bins for head pitch and eye angles for both full vision and blocked lower visual field conditions are shown in Fig. 4 for the smoothest surface (S1, top left) to the most complex (S14, bottom right).

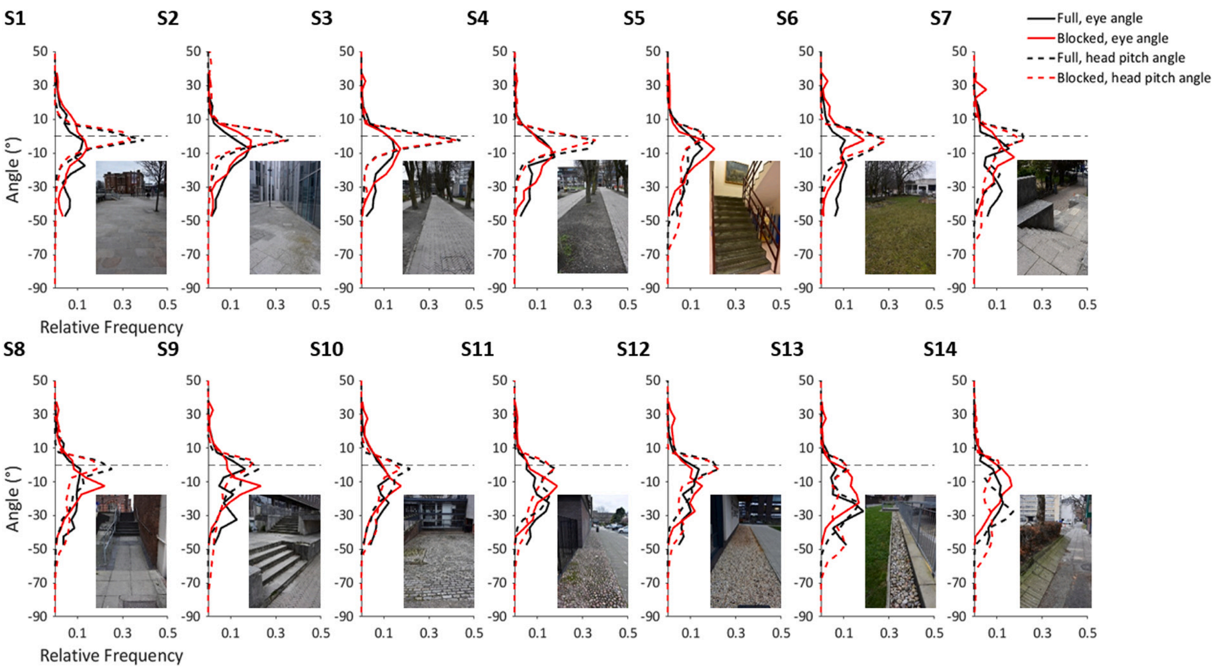

Fig. 4. Mean relative frequency distributions of head pitch (dashed line) and eye (solid line) angle for surfaces S1 to S14 under full vision (black) and blocked lower visual field (red) conditions. On the y-axis, results are plotted for 5° bins relative to 0° (looking straight ahead). Negative angles correspond to lowering of the eyes or head toward the ground. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

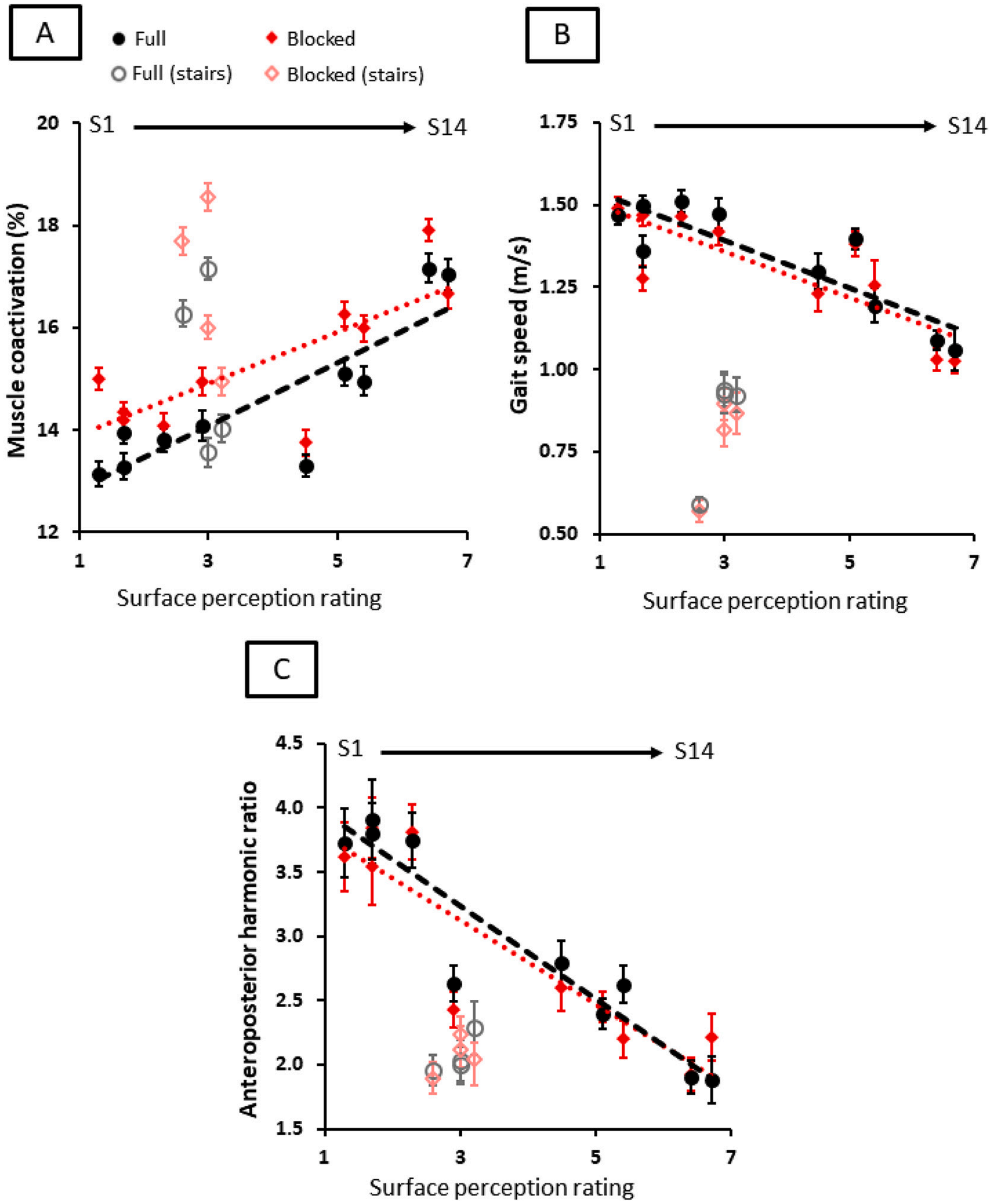

Fig. 5. Mean (± SE) (A) muscle coactivation, (B) gait speed and (C) anteroposterior harmonic ratio used to measure walking smoothness for surfaces S1 – S14 for full vision (black circles) and blocked lower visual field (red diamonds) conditions. Surfaces with stairs are represented separately (grey open circles and light red open diamonds respectively). Dotted lines represent the regression lines (when excluding stairs) for full vision and blocked lower visual field conditions. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

3.4. Muscle coactivation

Mean muscle coactivations are shown in Fig. 5A. A linear regression showed a significant relation between surface complexity and muscle coactivation for both full vision and blocked lower visual field conditions, (R^2 = 0.735; $F,(1,8)$ = 22.18, p = *0.002* and R^2 = 0.596; $F,(1,8)$ = 11.82, p = *0.009* respectively). However, there was no significant difference between the regressions intercepts or slopes (*t(19) = 1.39, p = 0.183* and *t(19) = 0.58, p = 0.573* respectively). Thus increasing surface complexity increased muscle coactivation to a similar extent for full vision and blocked lower visual field conditions.

3.5. Gait speed

Mean gait speeds are shown in Fig. 5B. A linear regression showed a significant relation between surface complexity and mean gait speed for both full vision and blocked lower visual field conditions, ($R^2 = 0.761$; $F,(1,8) = 25.524$, $p = 0.001$ and $R^2 = 0.688$; $F,(1,8) = 17.606$, $p = 0.003$ respectively). However, there was no significant difference between the two regression intercepts or slopes, ($t(19) = 0.41, p = 0.686$ and $t(19) = 0.09, p = 0.926$ respectively). Thus increasing surface complexity reduced gait speed to a similar extent for full vision and blocked lower visual field conditions.

3.6. Walking smoothness as measured by mean harmonic ratios

Mean anteroposterior harmonic ratios are shown in Fig. 5C. A linear regression showed a significant relation between surface complexity and walking smoothness from mean harmonic ratios for both full vision and blocked lower visual field conditions ($R^2 = 0.888$; $F,(1,8) = 63.29$, $p < 0.001$ and $R^2 = 0.813$; $F,(1,8) = 34.71$, $p < 0.001$ respectively). However, there was no significant difference between the two regression intercepts or slopes, ($t(19) = -0.68, p = 0.507$ & $t(19) = 0.48, p = 0.638$ respectively). Thus increasing surface complexity reduced harmonic ratios to a similar extent for full vision and blocked lower visual field conditions.

We also conducted a regression on the z-scores for each surface averaged over multiple measures to check if this provided a less noisy and more sensitive measure of the influence of blocking the lower visual field than the separate analyses of individual measures reported above. This was not found to be the case, see the supplementary material (SM5), with no significant difference between either the regression intercepts or slopes for full vision versus blocked lower visual field. Finally, Pearson's correlations were conducted on the mean z-scores of the different gaze and gait measures for each surface, first comparing full vision to blocked lower visual field (see supplementary materials Table SM6.1), then comparing the different measures for full vision (see supplementary materials Table SM6.2), and for blocked lower visual field (see supplementary materials Table SM6.3). For both visual conditions, head pitch and gait measures (muscle coactivation, gait speed and walking smoothness as measured by harmonic ratios) were more strongly correlated with each other than eye angle, fixation duration and number of fixations.

4. Discussion

Surfaces that were rated as rougher and less stable to walk on were associated with significant changes for both visual conditions (full vision and blocked lower visual field conditions). On more complex surfaces head pitch lowered, muscle coactivation increased, gait slowed and walking smoothness as measured by harmonic ratios was reduced. Thus surface complexity had wide-ranging effects on both gaze and gait. However, head pitch angle was the only measure that showed a significant synergistic interaction from the effect of blocking the lower visual field. The head lowered more when both the lower visual field was blocked and people walked over more complex surfaces compared to the sum of the individual effects of both factors, see Fig. 3A.

Gaze and gait behaviour showed clear differences when walking over more complex surfaces for both visual conditions. These results are in line with those reported in previous studies which have assessed gaze and gait across more challenging conditions ('t Hart & Einhauser, 2012; Marigold & Patla, 2007, 2008a; Matthis et al., 2018). In conjunction, these behavioural changes provide converging evidence that stability decreases when walking over more complex surfaces. As discussed in the introduction, given the current lack of a gold standard metric for stability (see Bruijn et al., 2013), using a diverse range and number of measures, as shown here, may provide a robust and sensitive indication of stability and therefore fall risk when walking.

Surface type had a greater impact on gaze and gait behaviour than blocking the lower visual field. Extrinsic factors, including the environment, are likely to be the predominant risk of falling in young healthy individuals (Berg & Cassells, 1990). In contrast, the elderly have intrinsic factors, affecting body functions, that increase their risk of falls (as reviewed in Pynoos, Steinman, & Nguyen, 2010). Understanding how vision affects stability is essential, given that elderly people with visual impairments report additional perceived risks of falling relative to elderly people without visual impairments (Brundle et al., 2015). In the present study the lack of behavioural changes when blocking the lower visual field, other than to head pitch and eye angle, suggests that young people are relatively robust to challenges to their locomotion. However, unlike in our study, Marigold and Patla (2008b) reported that gait speed reduced when the lower visual field was blocked. This discrepancy in findings may be because the surfaces used by Marigold and Patla (2008b) were more challenging given their multi-surface type composition. As such, this may indicate that young people need highly challenging conditions to reduce their stability. In support of this, studies that have included simulations of multiple intrinsic factors related to falling (e.g. both physical and cognitive impairments) have shown that young people adopt a cautious gait (Granacher, Wolf, Wehrle, Bridenbaugh, & Kressig, 2010; Hollman, Kovash, Kubik, & Linbo, 2007). Therefore, future studies investigating the factors influencing fall risk and walking stability in the elderly should either assess how young people cope with simulations of several concurrent impairments, representative of age-related co-morbidities, or directly assess behaviour indicative of stability in elderly populations.

Blocking the lower visual field only produced effects that interacted with surface complexity for one of our five measures, namely head pitch angle. This may indicate that we prioritise a relatively easy change (tilting the head down) over more energetically costly changes elsewhere in the body when walking on more complex surfaces. This would suggest that our initial tactic, when confronted with the challenge of walking on complex surfaces, is to improve the visual information available about the surfaces rather than altering our gait. Interestingly, previous research that has used goggles to block the lower visual field has not shown a lowering of head pitch in the young when stepping over obstacles (Muir, Haddad, Heijnen, & Rietdyk, 2015). Similarly multifocal spectacle wearers do not alter head pitch based on different lenses worn (Timmis, Johnson, Elliott, & Buckley, 2010). However, we assessed walking outside across complex surfaces, whereas this previous research focused on time-locked, short duration responses, including obstacle or step negotiation, performed predominately in laboratory based setting. Other research has found different responses to simulated visual deficits. For example, Zult, Allsop, Timmis, and Pardhan (2019) found that, when stepping over an obstacle with blurred vision, single stance support time as well as eye fixations increased compared to normal vision. Similarly, studies simulating monocular vision found that gait slowed and toe clearance increased when stepping over an obstacle, indicating a more cautious gait (Hayhoe, Gillam, Chajka, & Vecellio, 2009; Patla, Niechwiej, Racco, & Goodale, 2002). These differing results could indicate that lower visual field loss, as tested in the present study, has less effect on gait than other visual factors (e.g. blurred or monocular vision). However, alternatively, it may be that gait changes are more pronounced when people step over an obstacle rather than when they walk along a complex surface. Obstacle avoidance requires one-off adjustments to gait and, here, visual deficits may be more disruptive (Friedman et al., 2007; Jansen, Toet, & Werkhoven, 2010; Lajoie et al., 2018; Timmis & Buckley, 2012). As an example of this, Patla (1998) demonstrated that, when stepping over an obstacle, toe clearance increased and participants positioned their feet further from the obstacle when their lower visual field was blocked. In combination with findings at the head from Muir et al. (2015); Timmis et al. (2010), this suggests that when vision from the lower visual field is unavailable, eye and gait behaviour suffice to cope with immediate gait demands (obstacle negotiation), whereas head position is used to cope with long term challenges (e.g. uneven surfaces).

Changes in head pitch angle per se may influence the chance of falling in more challenging outside environments. This is because normally, with a flexed head position, people's gaze will centre on the ground plane surrounding their upcoming footsteps. This, in turn, means that people are less able to extract visual information from their wider surroundings, including street furniture, pedestrians and vehicles. A lowered head (typical of a flexed posture) is known to be associated with lower functional status (ability to perform normal daily activities) in elderly women (Balzini et al., 2003). Our test surfaces were located away from roads, crowds and static obstacles so this did not cause a problem for our participants. Nevertheless, such challenges are commonplace in everyday situations. The relative frequency plots (Fig. 4) in effect show the variance of eye and head pitch angles throughout the trial. Head pitch was nearly horizontal for much of the time spent walking on less complex, smoother surfaces under both full vision and blocked lower visual field. Here, participants could readily scan their surroundings. However, for the most complex surfaces, head pitch angle was both more variable and generally lower. In comparison, eye angle remained relatively constant under the two conditions regardless of surface complexity (see supplementary material SM7). Together these results suggest that participants were moving their head more when coping with more complex surfaces. Though not traditionally interpreted as a measure of stability, these changes to head pitch support the hypothesis that gaze angle (especially due to head movements) can be used to assess the complexity of a surface when walking (Thomas et al., 2020a).

One notable finding of the study was that stairs skewed most of our behavioural measures. Only for eye angle and eye fixation duration did the four surfaces with stairs (S5, S7, S8 and S9) produce behaviour similar to that of the other surfaces. The distinct behaviour for stairs was not surprising given previous research comparing walking up stairs to level walking (Cromwell & Wellmon, 2001; Wang et al., 2017; Zietz & Hollands, 2009). Stair walking has been researched with respect to gaze behaviour, biomechanics and muscle activity (e.g. Hinman, Cowan, Crossley, & Bennell, 2005; Miyasike-DaSilva, Allard, & McIlroy, 2011; Reeves, Spanjaard, Mohagheghi, Baltzopoulos, & Maganaris, 2008). Our results suggest that our surface complexity metric based on perceptual ratings may be inappropriate for stairs and that, instead, stairs might be better characterised based on physical measurements (Thomas et al., 2020b). We have, nevertheless, shown the results for stairs in both our figures and in the analyses reported in the supplementary materials, given that our aim was to test a broad range of surfaces typically encountered in everyday life. Furthermore it is important to understand the interaction between gaze and gait on stairs given that they are a common cause for falls, including for those with visual impairments (Pan, Liu, Sun, & Xu, 2015).

In summary, we found that many aspects of gaze and gait altered as surface complexity increased. In general, for our young, healthy participants the effects of surface complexity were not exacerbated by blocking the lower visual field. The only exception was for head pitch angle whereby the head lowered more on more complex surfaces if, in addition, the lower visual field was blocked. This finding illustrates the complexity of considering the effects of both extrinsic factors (e.g. surface complexity) and intrinsic factors (e.g. limiting visual information) on fall risk. Our study suggests that young people cope well with a reduction in information from their lower visual field even when walking over challenging surfaces. However, only one intrinsic factor was simulated here. Future research should see whether alternative or additional manipulations, representative of the comorbidities experienced by the elderly, change walking over surfaces of varying complexity. This would allow us to build up a more accurate understanding of how our gait responds to both extrinsic and intrinsic challenges.

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Author statement

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Declarations of Competing Interest

None.

Appendix A. Supplementary data

Supplementary data to this article can be found online at https://doi.org/10.1016/j.humov.2020.102676.

References

Alexander, M. S., Lajoie, K., Neima, D. R., Strath, R. A., Robinovitch, S. N., & Marigold, D. S. (2014). Effects of age-related macular degeneration and ambient light on curb negotiation. *Optometry and Vision Science, 91*, 975–989.
Balzini, L., Vannucchi, L., Benvenuti, F., Benucci, M., Monni, M., Cappozzo, A., & Stanhope, S. J. (2003). Clinical characteristics of flexed posture in elderly women. *Journal of the American Geriatrics Society, 51*, 1419–1426.
Bellanca, J., Lowry, K., VanSwearingen, J., Brach, J., & Redfern, M. (2013). Harmonic ratios: A quantification of step to step symmetry. *Journal of Biomechanics, 46*, 828–831.
Berg, R. L., & Cassells, J. S. (1990). *The second fifty years: Promoting health and preventing disability*: National Academy Press.
Beurskens, R., & Bock, O. (2012). Age-related decline of peripheral visual processing: The role of eye movements. *Experimental Brain Research, 217*, 117–124.
Black, A. A., Wood, J. M., & Lovie-Kitchin, J. E. (2011). Inferior field loss increases rate of falls in older adults with glaucoma. *Optometry and Vision Science, 88*, 1275–1282.
Bruijn, S., Meijer, O., Beek, P., & Van Dieën, J. (2013). Assessing the stability of human locomotion: A review of current measures. *Journal of the Royal Society Interface, 10*, 20120999.
Brundle, C., Waterman, H. A., Ballinger, C., Olleveant, N., Skelton, D. A., Stanford, P., & Todd, C. (2015). The causes of falls: Views of older people with visual impairment. *Health Expectations, 18*, 2021–2031.
Cheong, A. M., Geruschat, D. R., & Congdon, N. (2008). Traffic gap judgment in people with significant peripheral field loss. *Optometry and Vision Science, 85*, 26–36.
Chiacchiero, M., Dresely, B., Silva, U., DeLosReyes, R., & Vorik, B. (2010). The relationship between range of movement, flexibility, and balance in the elderly. *Topics in Geriatric Rehabilitation, 26*, 148–155.
Cohen, J. (2013). *Statistical power analysis for the behavioral sciences: Routledge*.
Collins, M. J., Brown, B., & Bowman, K. J. (1989). Peripheral visual acuity and age. *Ophthalmic and Physiological Optics, 9*, 314–316.
Crassini, B., Brown, B., & Bowman, K. (1988). Age-related changes in contrast sensitivity in central and peripheral retina. *Perception, 17*, 315–332.
Cromwell, R., & Wellmon, R. (2001). Sagittal plane head stabilization during level walking and ambulation on stairs. *Physiotherapy Research International, 6*, 179–192.
Foulsham, T., Walker, E., & Kingstone, A. (2011). The where, what and when of gaze allocation in the lab and the natural environment. *Vision Research, 51*, 1920–1931.
Franchak, J. M., & Adolph, K. E. (2010). Visually guided navigation: Head-mounted eye-tracking of natural locomotion in children and adults. *Vision Research, 50*, 2766–2774.
Friedman, D. S., Freeman, E., Munoz, B., Jampel, H. D., & West, S. K. (2007). Glaucoma and mobility performance: The Salisbury eye evaluation project. *Ophthalmology, 114*, 2232–2237 e2231.
Graci, V., Elliott, D. B., & Buckley, J. G. (2010). Utility of peripheral visual cues in planning and controlling adaptive gait. *Optometry and Vision Science, 87*, 21–27.
Granacher, U., Wolf, I., Wehrle, A., Bridenbaugh, S., & Kressig, R. W. (2010). Effects of muscle fatigue on gait characteristics under single and dual-task conditions in young and older adults. *Journal of Neuroengineering and Rehabilitation, 7*, 56.
Hayhoe, M., Gillam, B., Chajka, K., & Vecellio, E. (2009). The role of binocular vision in walking. *Visual Neuroscience, 26*, 73–80.
Hinman, R. S., Cowan, S. M., Crossley, K. M., & Bennell, K. L. (2005). Age-related changes in electromyographic quadriceps activity during stair descent. *Journal of Orthopaedic Research, 23*, 322–326.
Hollman, J. H., Kovash, F. M., Kubik, J. J., & Linbo, R. A. (2007). Age-related differences in spatiotemporal markers of gait stability during dual task walking. *Gait & Posture, 26*, 113–119.
Jansen, S. E., Toet, A., & Werkhoven, P. J. (2010). Obstacle crossing with lower visual field restriction: Shifts in strategy. *Journal of Motor Behavior, 43*, 55–62.
Kassner, M., Patera, W., & Bulling, A. (2014). Pupil: an open source platform for pervasive eye tracking and mobile gaze-based interaction. *Proceedings of the 2014 ACM international joint conference on pervasive and ubiquitous computing: Adjunct publication (pp. 1151-1160)*. ACM.
Lajoie, K., Miller, A. B., Strath, R. A., Neima, D. R., & Marigold, D. S. (2018). Glaucoma-related differences in gaze behavior when negotiating obstacles. *Translational Vision Science & Technology, 7*, 10.
Lee, W. J., Kim, J. H., Shin, Y. U., Hwang, S., & Lim, H. W. (2019). Differences in eye movement range based on age and gaze direction. *Eye, 1*.
Li, Q., Young, M., Naing, V., & Donelan, J. M. (2010). Walking speed estimation using a shank-mounted inertial measurement unit. *Journal of Biomechanics, 43*, 1640–1643.
Likert, R. (1932). *A technique for the measurement of attitudes.* (Archives of psychology).
Marigold, D. S., & Patla, A. E. (2007). Gaze fixation patterns for negotiating complex ground terrain. *Neuroscience, 144*, 302–313.
Marigold, D. S., & Patla, A. E. (2008a). Age-related changes in gait for multi-surface terrain. *Gait & Posture, 27*, 689–696.
Marigold, D. S., & Patla, A. E. (2008b). Visual information from the lower visual field is important for walking across multi-surface terrain. *Experimental Brain Research, 188*, 23–31.
Matthis, J. S., Barton, S. L., & Fajen, B. R. (2017). The critical phase for visual control of human walking over complex terrain. *Proceedings of the National Academy of Sciences of the United States of America, 114*, E6720–E6729.
Matthis, J. S., & Fajen, B. R. (2014). Visual control of foot placement when walking over complex terrain. *Journal of Experimental Psychology. Human Perception and Performance, 40*, 106–115.
Matthis, J. S., Yates, J. L., & Hayhoe, M. M. (2018). Gaze and the control of foot placement when walking in natural terrain. *Current Biology, 28*(1224–1233), Article e1225.
Menant, J. C., Steele, J. R., Menz, H. B., Munro, B. J., & Lord, S. R. (2009). Effects of walking surfaces and footwear on temporo-spatial gait parameters in young and older people. *Gait & Posture, 29*, 392–397.
Merryweather, A., Yoo, B., & Bloswick, D. (2011). Gait characteristics associated with trip-induced falls on level and sloped irregular surfaces. *Minerals, 1*, 109–121.
Mihailovic, A., Swenor, B. K., Friedman, D. S., West, S. K., Gitlin, L. N., & Ramulu, P. Y. (2017). Gait implications of visual field damage from glaucoma. *Translational Vision Science & Technology, 6*, 23.
Miller, A. B., Lajoie, K., Strath, R. A., Neima, D. R., & Marigold, D. S. (2018). Coordination of gaze behavior and foot placement during walking in persons with glaucoma. *Journal of Glaucoma, 27*, 55–63.
Miyasike-DaSilva, V., Allard, F., & McIlroy, W. E. (2011). Where do we look when we walk on stairs? Gaze behaviour on stairs, transitions, and handrails. *Experimental Brain Research, 209*, 73–83.
Muir, B., Haddad, J., Heijnen, M. J., & Rietdyk, S. (2015). Proactive gait strategies to mitigate risk of obstacle contact are more prevalent with advancing age. *Gait &*

Posture, 41, 233–239.
Pan, C.-W., Liu, H., Sun, H.-P., & Xu, Y. (2015). Increased difficulties in managing stairs in visually impaired older adults: a community-based survey. *PLoS One,* 10.
Patla, A. E. (1998). How is human gait controlled by vision. *Ecological Psychology, 10*, 287–302.
Patla, A. E., & Greig, M. (2006). Any way you look at it, successful obstacle negotiation needs visually guided on-line foot placement regulation during the approach phase. *Neuroscience Letters, 397*, 110–114.
Patla, A. E., Niechwiej, E., Racco, V., & Goodale, M. A. (2002). Understanding the contribution of binocular vision to the control of adaptive locomotion. *Experimental Brain Research, 142*, 551–561.
Patla, A. E., & Vickers, J. N. (1997). Where and when do we look as we approach and step over an obstacle in the travel path? *Neuroreport, 8*, 3661–3665.
Patla, A. E., & Vickers, J. N. (2003). How far ahead do we look when required to step on specific locations in the travel path during locomotion? *Experimental Brain Research, 148*, 133–138.
Pirker, W., & Katzenschlager, R. (2017). Gait disorders in adults and the elderly. *Wiener Klinische Wochenschrift, 129*, 81–95.
Pynoos, J., Steinman, B. A., & Nguyen, A. Q. (2010). Environmental assessment and modification as fall-prevention strategies for older adults. *Clinics in Geriatric Medicine, 26*, 633–644.
Reddy, R. S., & Alahmari, K. A. (2016). Effect of lower extremity stretching exercises on balance in geriatric population. *International Journal of Health Sciences, 10*, 389.
Reeves, N. D., Spanjaard, M., Mohagheghi, A. A., Baltzopoulos, V., & Maganaris, C. N. (2008). The demands of stair descent relative to maximum capacities in elderly and young adults. *Journal of Electromyography and Kinesiology, 18*, 218–227.
Rietdyk, S., & Drifmeyer, J. E. (2009). The rough-terrain problem: Accurate foot targeting as a function of visual information regarding target location. *Journal of Motor Behavior, 42*, 37–48.
t Hart, B. M., & Einhauser, W. (2012). Mind the step: Complementary effects of an implicit task on eye and head movements in real-life gaze allocation. *Experimental Brain Research, 223*, 233–249.
Thies, S. B., Richardson, J. K., & Ashton-Miller, J. A. (2005). Effects of surface irregularity and lighting on step variability during gait: A study in healthy young and older women. *Gait & Posture, 22*, 26–31.
Thomas, N. D. A., Gardiner, J. D., Crompton, R. H., & Lawson, R. (2020a). Look out: An exploratory study assessing how gaze (eye angle and head angle) and speed are influenced by surface irregularity. *PeerJ,* 8.
Thomas, N. D. A., Gardiner, J. D., Crompton, R. H., & Lawson, R. (2020b). Physical and perceptual measures of walking surface complexity strongly predict gait and gaze behaviour. *Human Movement Science, 71*.
Thompson, J. D., Plummer, P., & Franz, J. R. (2018). Age and falls history effects on antagonist leg muscle coactivation during walking with balance perturbations. *Clinical biomechanics, 59*, 94–100.
Timmis, M. A., Allsop, J., Baranian, M., Baker, J., Basevitch, I., Latham, K., ... van Paridon, K. N. (2017). Visual search behavior in individuals with retinitis pigmentosa during level walking and obstacle crossing. *Investigative Ophthalmology & Visual Science, 58*, 4737–4746.
Timmis, M. A., & Buckley, J. G. (2012). Obstacle crossing during locomotion: Visual exproprioceptive information is used in an online mode to update foot placement before the obstacle but not swing trajectory over it. *Gait & Posture, 36*, 160–162.
Timmis, M. A., Johnson, L., Elliott, D. B., & Buckley, J. G. (2010). Use of single-vision distance spectacles improves landing control during step descent in well-adapted multifocal lens-wearers. *Investigative Ophthalmology & Visual Science, 51*, 3903–3908.
Voloshina, A. S., Kuo, A. D., Daley, M. A., & Ferris, D. P. (2013). Biomechanics and energetics of walking on uneven terrain. *The Journal of Experimental Biology, 216*, 3963–3970.
Wang, K., Delbaere, K., Brodie, M. A., Lovell, N. H., Kark, L., Lord, S. R., & Redmond, S. J. (2017). Differences between gait on stairs and flat surfaces in relation to fall risk and future falls. *IEEE Journal of Biomedical and Health Informatics, 21*, 1479–1486.
Wang, K., Lovell, N. H., Del Rosario, M. B., Liu, Y., Wang, J., Narayanan, M. R., ... Lord, S. R. (2014). Inertial measurements of free-living activities: Assessing mobility to predict falls. *2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (pp. 6892-6895)*. IEEE.
Winter, D. A. (2005). *Biomechanics and motor control of human movement* (3rd ed.). Hoboken, New Jersey: John Wiley & Sons.
Zietz, D., & Hollands, M. (2009). Gaze behavior of young and older adults during stair walking. *Journal of Motor Behavior, 41*, 357–365.
Zult, T., Allsop, J., Timmis, M. A., & Pardhan, S. (2019). The effects of temporal pressure on obstacle negotiation and gaze behaviour in young adults with simulated vision loss. *Scientific Reports, 9*, 1–13.