Assessing Infant Carriage Systems: Ground Reaction Force Implications for Gait of the Caregiver

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Objective: To assess the acute alterations of anterior infant carriage systems on the ground reaction force experienced during over-ground walking.

Background: Previous research has identified the alterations in posture and gait associated with an increased anterior load (external or internal); however, the forces applied to the system due to the altered posture during over-ground walking have not been established.

Method: Thirteen mixed gender participants completed 45 over-ground walking trials at a self-selected pace under three loaded conditions (unloaded, semi-structured carrier 9.9 kg, and structured carrier 9.9 kg). Each trial consisted of a 15-m walkway, centered around a piezoelectric force platform sampling at 1,200 Hz. Differences were assessed between loaded and unloaded conditions and across carriers using paired samples *t* tests and repeated measures ANOVA.

Results: Additional load increased all ground reaction force parameters; however, the magnitude of force changes was influenced by carrier structure. The structured carrier displayed increased force magnitudes, a reduction in the time to vertical maximum heel contact, and an increased duration of the flat foot phase in walking gait.

Conclusion: Evidence suggests that the acute application of anterior infant carriers alters both kinetic and temporal measures of walking gait. Importantly, these changes appear to be governed not solely by the additional mass but also by the structure of the carrier.

Application: These findings indicate carrier structure should be considered by the wearer and may be used to inform policy in the recommendation of anterior infant carriage systems use by caregivers.

Keywords: biomechanics, gait, posture, kinetics, loading, product design

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HUMAN FACTORS
Vol. 60, No. 2, March 2018, pp. 160–171
DOI: 10.1177/0018720817744661

INTRODUCTION

The use of ergonomic aids, in the form of woven wraps, to assist in the transportation of infants has been and continues to be commonplace in developing countries throughout the world (Glover, 2012; Wu, Huang, & Wang, 2017). This approach has seen significant increase in developed countries over the past two decades (Frisbee & Hennes, 2000; Glover, 2012), resulting in the increased availability of commercial infant carriage devices. This trend can be in part attributed to its promotion by parenting organizations such as the National Childbirth Trust (2016), Babywearing International (2015), and the Centre for Babywearing Studies (2016). Proposed benefits include convenience, the promotion of physical development, child mental and physical health, safety, and improved health for the wearer (Natural Life Mom, 2012; Sling Babies, 2011). While some of these claims have been supported in the literature, including convenience (Wu et al., 2017), reduction in crying (Hunziker & Barr, 1986), and an increase in infant-mother attachment (Gathwala, Singh, & Balhara, 2008; Tessier et al., 1998), little attention has been directed toward the physical health of the caregiver, specifically, the short- and long-term implications of carrying an infant on the caregiver's posture, gait, and structural health.

The task of infant carriage is ostensibly one of load carriage, either anteriorly or posteriorly; however, the majority of load carriage work examine the effects of posterior load on posture (Atwells, Birrell, Hooper, & Mansfield, 2006; Schiffman, Bensel, Hasselquist, Gregorczyk, & Piscitelle, 2006), gait (Birrell & Halsam, 2008, 2010; Birrell, Hooper, & Haslam, 2007), ground reaction force (Birrell et al., 2007; Cavanagh & LaFortune, 1980; Ciacci, Di Michelea, & Mern, 2010; Hsiang, Jiang, & McGorry, 1998; Lloyd

& Cooke, 2000), fatigue (Qu & Yeo, 2011), and cardiovascular response (Fallowfield, Blacker, Willems, Davey, & Layden, 2012). Application of many of these findings are limited in reference to anterior load carriage given the significant differences reported by Fiolkowski, Horodyski, Bishop, Williams, and Stylianou (2006) in gait kinematics between anterior and posterior loads. However, findings associated with cardiovascular response, namely, the increased energy cost associated with an additional load, as measured by oxygen consumption and heart rate (Fallowfield et al., 2012), and increase in forces experienced proportionate to the load applied (Birrell et al., 2007) are more readily transferable. Consequently, the use of an anterior infant carriage system could have cardiorespiratory adaptations resulting in enhanced health and reduced disease risk, supporting the claims of parenting groups associated with the wearer's health (Natural Life Mom, 2012; Sling Babies, 2011). However, focus on anterior load carriage has been sparse in the academic literature with reference to posture and gait parameters (Birrell & Haslam, 2008; Fiolkowski et al., 2006; Graham, Smallman, Miller, & Stevenson, 2014; Hsiang et al., 1998; Junqueira, Amaral, Lutaka, & Duarte, 2015; Perry et al., 2010). Findings indicate that anterior load carriage, using a front pack equivalent to 10% and 15% of participant mass, caused a reduced hip flexion and extension compared to unloaded walking (Fiolkowski et al., 2006). While application of a fixed 4.4 kg load (divers belt) identified a significant decrease in vertical ground reaction force at maximum vertical thrust at push off (Birrell & Haslam, 2008), no other vertical force measures were significantly altered by the fixed anterior load. This may have been a result of the alteration in the center of gravity caused by the anterior mass, reducing the impulse needed to accommodate the load at push off (Hsiang et al., 1998). Furthermore, Junqueira et al. (2015) identified significant alterations in trunk orientation when participants carried live infants and infant mannequins in their arms. These were characterized by increased trunk inclination, lumbar lordosis, and thoracic kyphosis during standing posture and walking (Junqueira et al., 2015).

Associated literature addressing impact of increased anterior load can be found in analysis of gait during pregnancy; Junqueira et al. (2015) highlighted commonalities in qualitative movement patterns of pregnant gait and postpartum infant carriage gait. Furthermore, significant reduction in walking velocity during pregnancy (McCrory, Chambers, Daftary, & Redfern, 2011) and postpartum infant carriage (Junqueira et al., 2015) has been established. These alterations are suggested to be compensatory to mitigate the increased instability of the caregiver/child system caused by changes in the position of the center of gravity (Branco, Santos-Rocha, & Vieira, 2014). Consequently, these similarities may indicate that mothers are well positioned to transfer from in vivo carriage to postpartum carriage; however, nonmaternal caregivers will have had no such adaptations. Furthermore, the TICKS guidelines developed by the Consortium of UK Sling Manufacturers (National Childbirth Trust, 2016) state that the child should be positioned high on the chest, close enough for the carer to kiss the child on the forehead. In comparison to previous research, where loads were carried in a much lower position, the center of gravity will be raised, and therefore the alterations in gait characteristics further exaggerate.

Considering the alterations in walking posture and kinematics (Fiolkowski et al., 2006; Junqueira et al., 2015), understanding the loading of the body is important as joints and muscles will be loaded outside of the general motor pattern, exposing the wearer to increased prospects of injury (Bonci, 1999). This could be magnified by a lack of pregnancy adaptations in nonmaternal caregivers; therefore, the use of anterior infant carriage systems could have health implications for caregivers. In light of the limited work toward understanding the impact of anterior load, and specifically, that no research has yet established the impact of anterior infant carriage on the caregiver, the aim of the current research was to ascertain the changes in ground reaction forces experienced when carrying an anterior load on the chest using an infant carrier. Moreover, it aims to determine acute alterations in temporal and kinetic parameters of the foot ground interaction experienced by the caregiver during walking and if this is affected by specific carrier structure.

TABLE 1: Participant Demographics

Gender	*N* of Participants	Descriptive Statistic	Age (Years)	Height (cm)	Mass (kg)	BMI (kg/m^2)
Female	7	Minimum	23	162.30	57.60	19.80
		Maximum	48	174.60	90.70	31.24
		Mean	31.71	168.83	69.06	24.23
		SD	10.84	5.20	11.74	3.94
Male	6	Minimum	23	179	65.30	19.67
		Maximum	35	190.20	95	27.31
		Mean	26.5	184.37	80	23.49
		SD	4.59	4.29	11.67	2.90

METHOD

Participant Recruitment

Thirteen injury-free participants (female = 7; male = 6; mean age = 29.3 ± 8.65 years) volunteered to take part in this study. Participant demographics are presented in Table 1. This research complied with the tenets of the Declaration of Helsinki and was approved by the Institutional Review Board at Canterbury Christ Church University. Informed consent was obtained from each participant. Inclusion criteria required all participants to be free from injury at time of data collection and have had no back, lower limb, or shoulder injuries in the previous 12 months. No participants had given birth in the previous 12 months.

No previous research was deemed acceptable for an a priori sample size estimation; therefore, post hoc power analyses were conducted for the repeated measures *t* test and repeated measures ANOVA. Using G*Power (v. 3.19.2) with an alpha of 0.05 and a large effect size (0.8), demonstrated power was 0.75 and 0.9 for the *t* tests and ANOVA, respectively.

Experimental Conditions

Participants completed 15 barefoot trials at a self-selected walking pace over a 15-m distance in each condition (unloaded 1.54 ± 0.03 m·s^{-1}; and 2 anteriorly loaded conditions, structured [SC] 1.54 ± 0.03 m·s^{-1}; semi-structured [SSC] 1.52 ± 0.03 m·s^{-1}), making contact with their right foot on a force platform. Barefoot conditions were used to ensure that differences between shoe construction and condition between participants did not affect force measures as these have been demonstrated to influence force attenuation and foot and ankle kinematics during gait (Novacheck, 1998).

The unloaded condition was completed barefoot, wearing minimal clothing, defined as tight fitting top and sports shorts, and no carrier. The loaded conditions consisted of the wearing of two different anteriorly loaded infant carriers, one semi-structured and one structured, specifications for which can be found in Table 2. Both carriers were loaded with a purpose-made mannequin with the equivalent mass of a 12-month-old on the 50th percentile on the NHS (2017) growth charts (9.9 kg). A mannequin was used as it has previously been reported (Junqueira et al., 2015) that the carriage of a mannequin results in similar alterations in walking kinematics to carrying one's own infant when compared to unloaded walking. Participants were instructed to allow their arms to swing naturally during walking trials rather than holding on to the mannequin and carrier. The order in which the participants completed each condition was randomized using an online research randomizer (Urbaniak & Plous, 2013).

Instrumentation

Ground reaction force (GRF) data were sampled at 1,200 Hz via a 900 × 600 mm Kistler force platform (Model 9287BA, Kistler Instruments Ltd) using Bioware software (v5.3.0.7, Kistler Instruments Ltd). Timing lights (in house, Canterbury Christ Church University) recorded the time taken to complete the central 5 m of the overground walking trials, centered over the force platform, to allow calculation of walking velocity.

TABLE 2: Specifications of Infant Carriage Systems

Infant Carrier	Semi-Structured	Structured
Picture		
Weight (g)	576	997
Material	100% cotton	Main material: 60% cotton, 40% polyester Lining: 100% cotton Waist belt: 100% polyester Mesh: 100% polyester Cover for leg position zip: 100% cotton
Product features	A comfortable and supportive baby carrier that allows you to carry on your front, hip, or back Allows the carrier to grow with your child to fit any size of baby or toddler from newborn up to 4 years old Baby is securely supported in the best position for healthy hip development Wide shoulder straps to evenly spread the weight around your body and provide a custom fit for each user Padded waist for extra comfort	Ergonomic baby carrier with wide seat area Extra-padded shoulder straps Good stability in the waist belt Perfect for a newborn—no infant insert needed. Front-facing carrying option From newborn to 3 years Acknowledged as a hip-healthy baby carrier by International Hip Dysplasia Institute

Data Analysis

Data files containing GRF components for over-ground walking were filtered in Bioware (v5.3.0.7, Kistler Instruments Ltd) using a dual-pass Butterworth low-pass filter with a cutoff frequency of 50 Hz (McCrory, Chambers, Daftary, & Redfern, 2013; McCrory et al., 2011). A Fast Fourier Transformation of 13 randomly selected trials revealed data to be below 45 Hz. GRF data files were exported from Bioware to Excel, where a purpose-written analysis template extracted key kinetic and temporal components for further analysis. Peak vertical and anteroposterior force and impulse were calculated as key events in the loading of the gait cycle and have been demonstrated to be important responders to assess force during general load carriage (Birrell, Hooper, & Haslam, 2007). Rates of force loading and unloading were included to assess acceleration changes to the caregiver and carriage system, beyond that of maximal amplitudes, as indication of increased injury risk (Greenhalgh, Sinclair, Protheroe, & Chockalingham, 2012). The mediolateral assessment was included, as despite the small magnitude of these during walking gait, research suggests mediolateral stability is important in similar anterior load carriage tasks (Branco et al., 2013; Branco, Santos-Rocha, & Vieira, 2014; Lymbery & Gilleard, 2005) and therefore was deemed important for inclusion. Calculations for all variables are outlined in Tables 3 and 4, Figure 2, and with reference to Figure 1, except rate from peak medial force to max lateral force was calculated using maximums prior to midstance (MS). The fastest and the slowest walking trials were removed from analysis, leaving 13 trials per condition per participant, and GRF data were normalized to body weight (BW), and temporal measures were normalized to contact time.

TABLE 3: Kinetic Analysis

		Unloaded	Loaded (Combined)	Semi-Structured (SSC)	Structured (SC)
Peak force (BW)					
P_4	F_y – max posterior braking[a,b,c,d]	–0.250 ± 0.008	–0.272 ± 0.008	–0.269 ± 0.009	–0.275 ± 0.008
P_8	F_y – max anterior propulsive[a,b,c,d]	0.271 ± 0.007	0.307 ± 0.008	0.304 ± 0.008	0.309 ± 0.009
P_3	F_x – medial peak force[a,b,d]	0.068 ± 0.007	0.075 ± 0.008	0.074 ± 0.008	0.076 ± 0.008
Impulses (BW·s)					
I_1	F_x – medial impulse	0.0020 ± 0.0002	0.0022 ± 0.0002	0.0022 ± 0.0002	0.0022 ± 0.0002
I_2	F_y – braking impulse[a,b,c,d]	–0.0371 ± 0.0011	–0.0420 ± 0.0013	–0.0418 ± 0.0015	–0.0422 ± 0.0011
I_3	F_y – propulsive impulse[a,b,c,d]	0.0358 ± 0.0013	0.0406 ± 0.0013	0.0404 ± 0.0013	0.0409 ± 0.0013
Loading rates ($BW \cdot s^{-1}$)					
$\frac{Fz_{P_2} - Fz_{P_1}}{t_{P_2} - t_{P_1}}$	F_z – impact loading rate[a,b,c,d]	51.285 ± 3.430	56.256 ± 3.680	55.678 ± 3.687	56.841 ± 3.746
$\frac{F_{zP_9} - Fz_{P_7}}{t_{P_9} - t_{P_7}}$	F_z – load off rate[a,b,c,d]	–16.615 ± 0.588	–18.725 ± 0.718	–18.413 ± 0.706	–19.038 ± 0.763
$\frac{Fx_{P_3} - Fx_{P_1}}{t_{P_3} - t_{P_1}}$	F_x – medial impact loading rate[a,b,d]	3.184 ± 0.458	3.681 ± 0.577	3.632 ± 0.547	3.730 ± 0.611
—	F_x - max med. to max lat. rate[b,c]	1.257 ± 0.235	1.428 ± 0.216	1.450 ± 0.207	1.406 ± 0.232
$\frac{Fz_{P_6} - Fz_{P_5}}{t_{P_6} - t_{P_5}}$	F_z – MHC to MS load off rate[b,d,e]	–3.436 ± 0.298	–3.654 ± 0.299	–3.525 ± 0.304	–3.784 ± 0.307
$\frac{Fz_{P_7} - Fz_{P_6}}{t_{P_7} - t_{P_6}}$	F_z – MS to MaxT load rate	2.658 ± 0.164	2.734 ± 0.145	2.716 ± 0.156	2.751 ± 0.141
$\frac{Fy_{P_4} - Fy_{P_1}}{t_{P_4} - t_{P_1}}$	F_y – braking force rate	–3.681 ± 0.379	–4.064 ± 0.493	–3.950 ± 0.477	–4.178 ± 0.516
Delta changes (BW)					
$P_6 - P_5$	F_z – MHC – MS difference[a,b,d]	0.543 ± 0.050	0.603 ± 0.050	0.585 ± 0.051	0.622 ± 0.050
$P_5 - P_7$	F_z – MHC – MaxT difference[a,b,d]	0.057 ± 0.034	0.100 ± 0.038	0.089 ± 0.037	0.112 ± 0.039
$P_7 - P_6$	F_z – MS – MaxT difference	0.486 ± 0.025	0.503 ± 0.023	0.496 ± 0.025	0.510 ± 0.022

Note. BW = body weights; MHC = maximum heel contact; MS = midstance; MaxT = maximum thrust.
[a]Denotes significant difference between loaded and unloaded condition.
[b]Denotes a significant finding from repeated measures ANOVA.
[c]Denotes significant pairwise comparison between unloaded and semi-structured.
[d]Denotes significant pairwise comparison between unloaded and structured.
[e]Denotes significant pairwise comparison between semi-structured and structured.

Statistical Analysis

Data were checked for normality using a Shapiro-Wilk test. Pairwise assessment of loaded (combination of the two loaded conditions) versus unloaded and between-carrier conditions were conducted to ascertain if carrier structure was associated with any significant differences in GRF. Pairwise comparison between loaded and unloaded were calculated using either a paired *t* test or a Wilcoxon matched pairs test, from which a Cohen's *d* effect size was calculated and interpreted as small (0.2), medium (0.5), and large (0.8). Cross-carrier assessment employed either a repeated measures ANOVA across carrier condition, where significant differences were ascertained through pairwise comparison using a Bonferroni post hoc analysis or a Friedman test. Effect size for the repeated measures ANOVA was calculated using partial eta-square and interpreted as small (0.01), medium (0.09), and large (0.25). The alpha level for all tests was set to 0.05, and all tests were carried out using SPSS (version 22, IBM).

RESULTS

Unloaded Versus Loaded

Kinetic analysis. Pairwise comparisons between unloaded (Un) and loaded conditions (Table 3) identified significant increases in impact force peak (IFP): $t(10) = -3.243$, $p = .009$, $d = 0.98$, Un: 0.745 ± 0.034, loaded: 0.869 ± 0.054; maximum heel contact (MHC): $t(12) = -11.307$, $p = .000$, $d = 3.14$, Un: 1.206 ± 0.032, loaded: 1.375 ± 0.034; midstance (MS): $t(12) = -10.752$, $p = .000$, $d = 2.98$, Un: 0.663 ± 0.019, loaded: 0.772 ± 0.019; and maximum vertical thrust (vertical force component propulsive peak; MaxT): t(12) = 14.714, p = .000, d = 4.08, Un: 1.149 ± 0.012, loaded: 1.275 ± 0.017, under loaded conditions. Rate of vertical force loading at heel contact (VLR) and rate of force unloading (VLOR) at end of stance also significantly increased when loaded, $t(12) = -3.890$, $p = .002$, $d = 1.08$; $t(12) = 6.283$, $p = .000$, $d = 1.74$, respectively.

Similarly, maximum posterior braking force (MPB), $t(12) = 4.566$, $p = .001$, $d = 1.27$, and maximum anterior propulsive force (MAP), $t(12) = -6.734$, $p = .000$, $d = 1.87$, significantly increased under loaded conditions. Loaded walking also resulted in significant increases in braking impulse (BI) and propulsive impulse (PI), $t(12) = 8.921$, $p = .000$, $d = 2.47$; $t(12) = -7.852$, $p = .000$, $d = 2.18$, respectively; however, rate of braking force (BLR) application was not significantly altered ($z = -1.645$, $p = .101$). Furthermore, load significantly increased the medial peak force (MPF), $t(12) = -2.386$, $p = .034$, $d = .66$, and medial loading rate (MLR), $z = -2.481$, $p = .013$, $d = 0.69$; however, medial impulse (MI) did not alter significantly.

In the transition from MHC to MaxT (the flat foot phase of stance), significant increases were evident in the magnitude of force changes between MHC and MS, $t(12) = -2.812$, $p = .016$, $d = 0.78$, and MHC and MaxT, $t(12) = -3.156$, $p = .008$, $d = 0.88$, under loaded conditions. However, these changes were not sufficient to significantly alter the load-off rate between MHC and MS and the rate of force application from MS to MaxT ($p > .05$) or the force magnitude change between MS and MaxT.

Temporal analysis. A significant, $t(12) = -2.260$, $p = .043$, $d = -0.63$, increase in time between the MHC and MaxT in the loaded condition (57.81 ± 0.57%) in comparison to unloaded (56.91 ± 0.47%) indicated that the load increased the time during which participant's full foot was in contact with the force platform. No significant difference between loaded/unloaded conditions were observed for other temporal measures (Table 4).

Analysis by Carrier Type

Kinetic analysis. Significant findings in all vertical force measures from the paired samples *t* tests were duplicated in the overall effect of the repeated measures ANOVA when load was separated by carrier type. Post hoc pairwise comparisons identified the magnitude of the MHC, $F(2, 24) = 96.589$, $p < .001$, $\eta^2 = 0.89$, was significantly higher under the SC condition (1.390 ± 0.034 BW), with MHC diminishing through the SSC condition (1.361 ± 0.035 BW) to the unloaded condition (1.206 ± 0.032 BW). All other vertical force measures exhibited no difference between SSC and SC, and this held true for the peak forces and impulses in anterior posterior forces. Medial impulse, $F(1.359, 16.309) = 1.9$,

TABLE 4: Temporal Analysis

Calculation		Unloaded	Loaded (Combined)	Semi-Structured	Structured
$t_{P_9} - t_{P_1}$	Contact time (s)	0.604 ± 0.009	0.607 ± 0.008	0.608 ± 0.008	0.606 ± 0.009
$\frac{t_{P_2} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to impact peak (% CT)	3.95 ± 0.55	4.26 ± 0.91	4.49 ± 1.08	4.02 ± 0.80
$\frac{t_{P_5} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to max heel contact (% CT)[a]	21.29 ± 0.44	20.65 ± 0.46	20.87 ± 0.53	20.43 ± 0.42
$\frac{t_{P_6} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to midstance (% CT)	47.34 ± 0.61	47.87 ± 0.71	48.11 ± 0.74	47.64 ± 0.71
$\frac{t_{P_6} - t_{P_5}}{t_{P_9} - t_{P_1}}$	Time to MS from MHC (% CT)	26.05 ± 0.63	27.22 ± 0.72	27.23 ± 0.72	27.21 ± 0.72
$\frac{t_{P_7} - t_{P_1}}{t_{P_9} - t_{P_1}}$	Time to MaxT (% CT)	78.20 ± 0.43	78.46 ± 0.45	78.38 ± 0.48	78.53 ± 0.43
$\frac{t_{P_7} - t_{P_5}}{t_{P_9} - t_{P_1}}$	Time to MT from MHC (% CT)[a,b,c]	56.91 ± 0.47	57.81 ± 0.57	57.51 ± 0.61	58.11 ± 0.56
$\frac{t_{P_7} - t_{P_6}}{t_{P_9} - t_{P_1}}$	Time to MaxT from MS (% CT)	30.62 ± 0.57	30.55 ± 0.55	30.25 ± 0.55	30.86 ± 0.61
$\frac{t_{P_9} - t_{P_7}}{t_{P_9} - t_{P_1}}$	Time from MaxT to toe off (% CT)	21.80 ± 0.43	21.54 ± 0.45	21.62 ± 0.48	21.47 ± 0.43
	Time max medial force to max lateral force (%CT)	20.64 ± 1.92	19.45 ± 1.60	19.35 ± 1.77	19.56 ± 1.55
—	Velocity final ($m \cdot s^{-1}$)	1.538 ± 0.031	1.532 ± 0.031	1.519 ± 0.030	1.545 ± 0.032

Note. % CT = percentage of contact time; MS = midstance; MHC = maximum heel contact; MaxT = maximum thrust.
[a]Denotes a significant finding from repeated measures ANOVA.
[b]Denotes significant difference between loaded and unloaded condition.
[c]Denotes significant pairwise comparison between unloaded and semi-structured.

$p = .171$, $\eta^2 = 0.137$, and MPF, $F(1.399, 16.794) = 4.91$, $p = .031$, $\eta^2 = 0.29$, also echoed the paired samples analysis; however, Bonferroni post hoc pairwise analysis of the MPF was too conservative to identify the source of the significant difference. Further investigation using a lowest significant difference (equivalent to no adjustments) identified the source of the difference, with the SC condition (0.076 ± 0.008 BW) MFP magnitude being significantly higher than the unloaded condition (0.068 ± 0.007 BW). Resultantly, the same pattern of significance was identified in the MLR, $\chi^2(2) = 6.615$, $p = .037$, and the Wilcoxon post hoc identified significance between SC and unloaded ($z = -2.481$, $p = .013$, $d = 0.69$).

The transition from MHC to MaxT highlighted further impacts of the SC carrier condition. The magnitude change between MHC and MS, $F(2, 24) = 7.267$, $p = .003$, $\eta^2 = 0.38$, was significantly impacted by load condition, and post hoc testing highlighted a significantly greater drop from MHC to MS (SC: 0.622 ± 0.050, Un: 0.543 ± 0.050). This led to a significant increase in the load-off rate from MHC to MS, $F(2, 24) = 3.707$, $p = .040$, $\eta^2 = 0.236$.

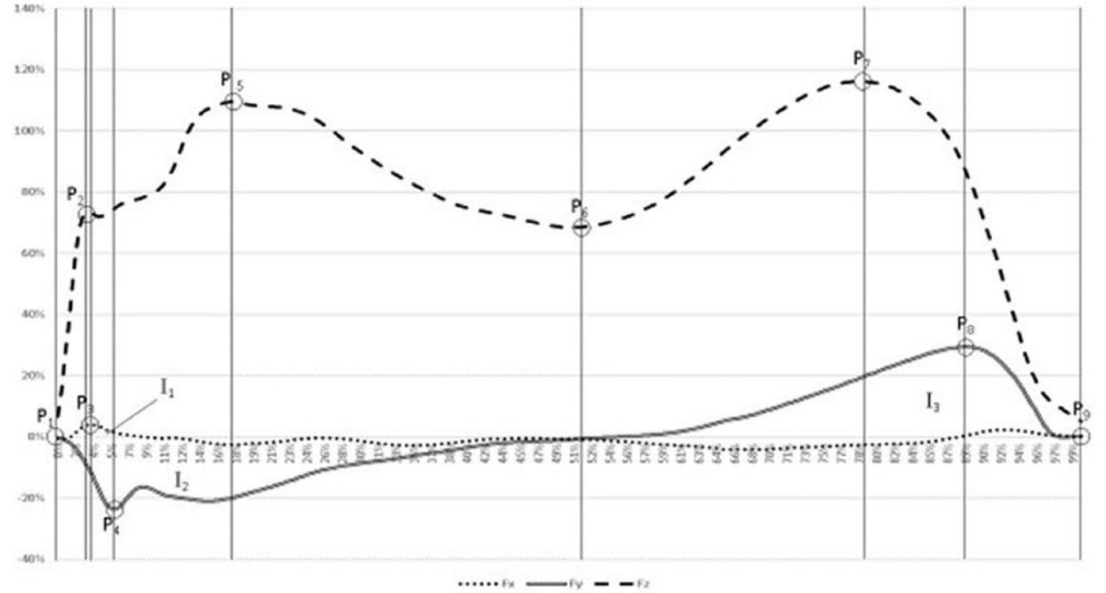

Figure 1. Annotated typical ground reaction force trace. P1 – initial contact, P2 – impact force peak, P3 – medial peak force, P4 – max posterior braking, P5 – max heel contact, P6 – midstance, P7 – max vertical thrust, P8 – max anterior propulsive, P9 – toe off, I1 – medial impulse, I2 – braking impulse, I3 – propulsive impulse. Fx = mediolateral force; Fy = anterior posterior force; Fz = vertical force.

Interestingly, the SC condition (–3.784 ± 0.307 $BW.s^{-1}$) was significantly faster than the unloaded (–3.436 ± 0.298 $BW.s^{-1}$) but not the SSC (–3.525 ± 0.304 $BW.s^{-1}$, p = .051) conditions, although this finding required the use of the least significant differences approach due to the conservative nature of the Bonferroni post hoc previously identified.

The magnitude of change between MHC and MaxT showed further significance alterations due to increased anterior load, $F(2, 24) = 8.201$, $p = .002$, $\eta^2 = 0.406$; again, the SC condition (0.112 ± 0.039 BW) was significantly higher than unloaded (0.057 ± 0.034 BW).

Temporal analysis. A significant difference between conditions for the time between MHC and MaxT as a percentage of contact time, $F(2, 24) = 5.152$, $p = .014$, $\eta^2 = 0.30$, was apparent. Post hoc pairwise comparisons identified a significant difference between unloaded (56.91 ± 0.47% CT) and the SC (58.11 ± 0.56% CT) condition; the SSC condition exhibited no significant difference (57.51 ± 0.61% CT). While the ANOVA for time from heel contact to MHC indicated significant differences, $F(2, 24) = 3.475$, $p = .047$, $\eta^2 = 0.23$, Bonferroni post hoc tests were too conservative to identify the specific source of the difference reported. Further examination using a lowest significant difference (equivalent to no adjustments) post hoc assessment identified the significant difference between Un (21.29 ± 0.44% CT) and SC (20.43 ± 0.42% CT) conditions (p = .040). Examination of the data indicates that both the loaded conditions (SSC 20.87 ± 0.53% CT; SC 20.43 ± 0.42% CT) were characterized by a faster move to MHC than the Un condition (21.29 ± 0.44% CT), although not enough to significantly alter loading rate at impact.

DISCUSSION

Previous studies have identified a decrease in walking velocity in response to the addition of an external anterior load (Junqueira et al., 2015) and internal anterior mass (McCrory et al., 2011). Our findings do not support this. While this contradiction was unexpected, previous work used mothers only and had the infant (or mannequin) supported in the arms; conversely, the focus of this work was to investigate the

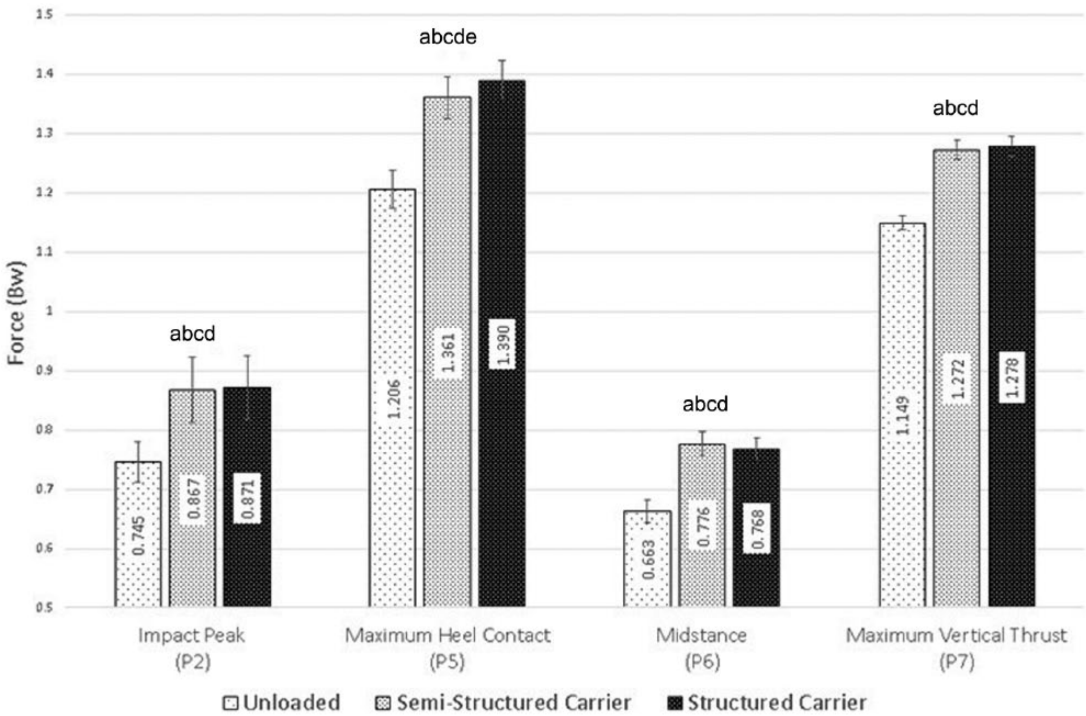

Figure 2. Vertical force parameter changes due to load and carrier type.
[a]Denotes significant difference between loaded and unloaded condition.
[b]Denotes a significant finding from repeated measures ANOVA, indicating difference between the three conditions.
[c]Denotes significant pairwise comparison between unloaded and semi-structured.
[d]Denotes significant pairwise comparison between unloaded and structured.
[e]Denotes significant pairwise comparison between semi-structured and structured.

caregiver (non–gender specific) and investigated the effect of an ergonomic aid (carrier) to assist in the carriage task.

Statistically significant increases in peak vertical force parameters (IP, MHC, MS, MaxT) were demonstrated in both loaded conditions when compared with unloaded (Figure 2). These increases were in direct opposition to Birrell and Haslam (2008), who found a significant reduction in MaxT in response to the load, with all other measures demonstrating no change. This contradiction is likely due to the increased load used in the current study (9.9 kg vs. 4.4 kg, Birrell & Haslam, 2008) and could be influenced by the raised position of the anterior mass. Interestingly, the significant increase in MaxT combined with the significant increase in propulsive impulse, $t(12) = 8.921$, $p = .000$, $d = 2.47$, also contradicted the propositions of Hsiang et al. (1998), who suggested a decreased impulse was required at push off with an anterior load. These contradictions could have resulted from the nature of the task as previous research had addressed a load (Birrell & Haslam, 2008; Hsiang et al., 1998) and the current research was that of carrying an infant. While this proposition cannot be proved due to the psychological influences being outside of the scope of this paper, participants in the current study may have used an external focus of attention and as such altered the gait accordingly. It has been shown that this approach can affect motor patterns (Wulf, Weigelt, Poulter, & McNevin, 2003), and Junqueria et al. (2015) identified that mannequin carriage and infant carriage demonstrate similar variations from unloaded walking.

Increases early in foot contact, specifically the MHC, displayed variation beyond that of the load alone. The use of a structured carrier resulted in a significantly higher force (SC: 1.390 ± 0.034 BW) being experienced by the caregiver compared to the SSC and unloaded conditions (SSC: 1.361 ± 0.035 BW, Un: 1.206 ± 0.032 BW), which was deemed to have a large effect (η^2 = 0.89). This may have been influenced by the significant reduction in time between heel contact and MHC, $F(2, 24) = 3.475$, $p = .047$, $\eta^2 = 0.225$. Post hoc assessment identified significance values of 0.070 between SC and SSC and 0.040 between SC and unloaded, indicating that the use of the SC is characterized by a quicker transfer from heel contact to MHC through increased acceleration, therefore resulting in the increased force measured at MHC. When considered alongside the findings of Fiolkowski et al. (2006) and Junqueira et al. (2015), both of whom applied loads of similar magnitude to the current study (15% and 10 kg, respectively), and Bonci (1999), this indicates that the caregiver is being exposing to increased stresses and possibly enhanced risk of injury due to greater magnitude of all vertical forces, reduced time to maximum load (MHC), and the alterations in posture previously described.

The significantly higher MHC peak force resulted in further significant alterations in the loading patterns during walking. The magnitude of the reduction from MHC to MS was significantly larger under loaded conditions, $t(12) = –2.812$, $p = .016$, $d = 0.78$. Analysis by carrier identified statistical significance between SC and unloaded ($p = .022$). This linked to further significant differences between loaded and unloaded, $F(2, 24) =3.707$, $p = .040$, $\eta^2 = 0.24$, with the rate of force unloading from MHC to MS being significantly faster under SC conditions when compared to the unloaded condition (SSC: $p = .051$), placing the caregiver under greater extremes of force. As a result, this may increase the likelihood of injury, if the carrier were employed for prolonged use, or could positively affect the caregiver through the overload principle attributed to resistance training (Winett & Carpinelli, 2001). Further research addressing prolonged use is needed to ascertain the veracity of these propositions.

The combination of the decrease in time from heel contact to MHC and maintenance of the contact time, $t(12) = –0.558$, $p = .587$, resulted in a significant increase in duration, as a percentage of contact time, between MHC to MaxT, with participants spending significantly longer in this transition when wearing the infant carriage systems. When carriage systems were separated, $F(2, 24) = 5.152$, $p = .014$, $\eta^2 = 0.30$, the foundation of this increase was the SC, exhibiting an increase of 1.2%. This increase in time indicates a longer period of knee flexion during stance, requiring greater muscular effort, especially given the increased forces experienced under loaded conditions. Furthermore, increasing the time in the flat foot phase of stance may be indicative of participants attempting to stabilize the system. It has been reported through kinematic analysis that stability of the body is the primary focus of the pregnant woman during gait (Branco et al., 2013, 2014; Lymbery & Gilleard, 2005), and thus it could be expected that the same would be true of postpartum mothers and other caregivers when carrying an infant in a carrier system. While little previous research has identified kinetic changes (Lymbery & Gilleard, 2005), what has been identified supports the importance of stability. Although the temporal findings do not directly speak to previous research addressing stability, as emphasis has been placed on the importance of stability in the medio-lateral direction (Branco et al., 2013, 2014; Lymbery & Gilleard, 2005), they do indicate that an acute adaptation has been employed by participants resulting in a longer duration of flat foot contact and greater stability.

Analysis of the medio-lateral parameters from both the pairwise analyses and repeated measures ANOVA displayed mixed findings. Alterations in medio-lateral parameters were inconsistent; the medial impulse demonstrated no significant alteration due to load or carrier type, where rate of force transfer from medial to lateral peak, MPF, and MLR all displayed significant alterations in response to load. The rate of force transfer from medial to lateral during stance, although significant overall, $\chi^2(2) = 6.000$, $p = .050$, did not clearly demonstrate the specific source as post hoc analysis, using alpha level correction, and could not ascertain the specific source of the significance. The MPF and consequently MLR increases, $t(12) = -2.386$, $p = .034$, $d = 0.66$; $Z = -2.481$, $p = .013$, $d = 0.69$, respectively, were solely a function of the SC condition. These findings, combined with those from the temporal and vertical force analysis, indicate that carrier structure in addition to load has an influence on the magnitude of forces experienced by the wearer. Careful consideration is therefore required when selecting an anterior infant carrier.

CONCLUSION

Results indicate that the use of an infant carrier caused a significant increase in the magnitudes of the forces experienced during walking and altered the temporal characteristics of caregiver gait. The significant increases in the ground reaction forces are largely a result of the increased load applied to the system, with increases in both magnitude and rate of force application influenced. However, the localized changes due to carrier type in both kinetics and temporal measures indicate that carrier structure has an influence beyond the magnitude of the load. Resultantly, caregivers should be cautious when selecting and using such devices as these results are based on acute application only, without consideration of prolonged use. Further investigation would be merited in exploring the postural changes associated with the observed alterations in ground reaction force and the impact of prolonged use on the wearer.

KEY POINTS

- Carrying infants in ergonomic carriers has been said to improve the bond between caregiver and child, but the implications of wearer health have received limited attention.
- Previous work has addressed maternal mothers only with no consideration for other caregivers in an infant's life.
- Results indicate that the load increases all aspects of ground reaction force; however, the magnitude and temporal alterations are dependent on carrier structure.
- The structure of the carrier should be carefully considered when selecting ergonomic infant carriers as the acute alterations indicate that the structure impacts the magnitude of the forces experienced during over-ground walking.

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Date received: May 11, 2017
Date accepted: September 14, 2017