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Sex Differences in Head Acceleration During Heading While Wearing Soccer Headgear

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Context: Researchers have indicated that female soccer players may be at greater risk of concussion compared with their male counterparts. Soccer headgear is marketed for reducing head acceleration and risk of concussion.

Objective: To determine the effect of sex and soccer headgear on head impact kinematics and dynamic stabilization during soccer heading.

Design: Cross-sectional design.

Setting: Research laboratory.

Patients or Other Participants: Forty-four college-aged soccer players (29 women, 15 men).

Intervention(s): Using a head impact model, participants performed 4 soccer headers under 3 headgear conditions (control, Head Blast Soccer Band, and Full90 Select Performance Headguard).

Main Outcome Measure(s): Dependent variables assessed before soccer heading were head-neck anthropometrics and isometric neck muscle strength, and those assessed during soccer headers were resultant linear head acceleration, Head Injury Criteria (HIC_{36}), and superficial neck muscle electromyography. Statistical analyses included multivariate and univariate analyses of variance with repeated measures, independent-samples *t* tests, appropriate follow-up analyses of variance and post hoc *t* tests, and Pearson product moment correlations ($\alpha = .05$).

Results: Head acceleration in women was 32% and 44% greater than in men when wearing the Head Blast (21.5 *g* versus 16.3 *g*) and Full90 Select (21.8 *g* versus 15.2 *g*), respectively ($P < .05$). Compared with men, women exhibited 10% greater head accelerations (20.2 *g* versus 18.2 *g*) during the control condition ($P = .164$).

Conclusions: Female soccer players exhibited greater head accelerations than their male counterparts when wearing headgear. Our results are important clinically because they indicate that soccer headgear may not be an appropriate head injury prevention tool for all athletes.

Key Words: head impact kinematics, concussion pathomechanics, brain injuries, football players

Key Points

- Women exhibited greater head impact accelerations compared with men when wearing headgear.
- Soccer headgear may not be an appropriate head injury prevention tool for all athletes.

Researchers[1–4] have reported that female soccer players have a greater incidence of concussion compared with male players. This finding may be due to differences in head mass[5] and neck muscle strength (ie, stability qualities).[6] Based on the Newton Second Law of Motion (F = ma), less head mass and neck strength should result in greater head accelerations upon force application. Head impact acceleration is important to limit because it is directly related to brain injury.[7] Using a nonfunctional pulley testing system, researchers have reported greater head accelerations in physically active women than in physically active men.[6] Sex differences in head acceleration among college athletes have not been reported.[8]

Soccer headgear has been developed to reduce the risk that an athlete will sustain head injury by reducing impact acceleration. Laboratory research on the effect of soccer headgear has been limited to using head forms,[9–13] force plates,[14] drop tests,[11–13] or a small sample of human participants.[10] Headgear seems effective in reducing impact force and the resultant acceleration at high ball speeds (≥34 mph [15.20 m/s]) and stiff impacts (eg, head forms, force plates). Use of a reliable head impact model involving controlled soccer heading could allow investigators to explore the mechanisms underlying individual kinematic response variations to head impact.[15]

The purpose of our study was to assess the effect of sex and headgear on head impact kinematics and dynamic stabilization during soccer heading in college-aged, experienced soccer players. We hypothesized that female stability qualities would be lower and would yield greater head impact kinematics compared with male stability qualities. We also hypothesized that the soccer headgear would exhibit no effect on head impact kinematics (ie, no main or interaction effects).

METHODS

Design

The study consisted of a cross-sectional research design. The independent variables were sex (female, male) and headgear (control, Head Blast Soccer Band [Head

Figure 1. Soccer headgear. A, Full90 Select Performance Headguard (Full90 Sports Inc, San Diego, CA). B, Head Blast Soccer Band (Head Blast Soccer Band Co, St Louis, MO).

Blast Soccer Band Co, St Louis, MO] and Full90 Select Performance Headguard [Full90 Sports Inc, San Diego, CA]) (Figure 1). Headgear was a repeated measure and a randomized factor. The dependent variables assessed before soccer heading included head-neck segment anthropometrics and isometric neck muscle strength. Resultant linear head acceleration measured in gravitational units, Head Injury Criteria (HIC_{36}), preparatory and reactive peak (percentage of maximal voluntary isometric contraction [% MVIC]), and area (% MVIC $\times$ ms) muscle activity of the right and left sternocleidomastoid and upper trapezius muscles were assessed during soccer heading.

Resultant linear head acceleration was defined as the greatest digitized point of acceleration after impact and was measured using a triaxial mouthpiece accelerometer. Time of impact was determined using a force-sensitive resistor (FSR) placed on the center of the forehead and secured with self-adhesive tape. Peak muscle amplitude was defined as the highest amplitude during 1 trial. Muscle amplitude area was defined as the sum of the amplitudes of activity multiplied by the total time of the trial. Peak MVIC values were determined during the neck flexor and extensor isometric strength test in each participant, and the peak muscle amplitude and muscle area were normalized to these values.

Time histories of the resultant linear accelerations were exported from the TeleMyo system (Noraxon USA, Scottsdale, AZ) to an Excel (Office 2003; Microsoft Corp, Redmond, WA) spreadsheet (Figure 2). Data then were exported to the HIC software (National Highway Traffic Safety Administration, US Department of Transportation, Washington, DC), and HIC_{36} was determined based on the following equation:

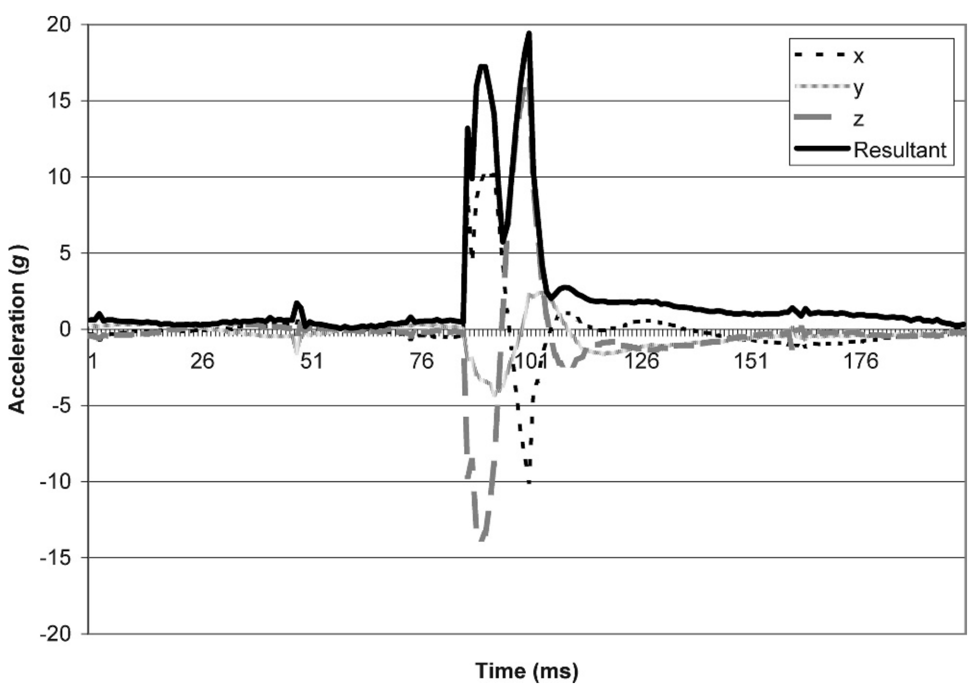

Figure 2. Examples of acceleration-time curves.

$$HIC = \max_{t_1,t_2}\left\{(t_2-t_1)\left[\frac{\int_{t_1}^{t_2}\bar{a}(t)dt}{t_2-t_1}\right]^{2.5}\right\}$$

where $\bar{a}(t)=\sqrt{a_x(t)^2+a_y(t)^2+a_z(t)^2}$ is the resultant acceleration in gravitational units versus time in seconds and where $t_2 - t_1 \leq 36$ milliseconds is the time span that maximizes the results of the HIC.

Participants

Forty-four volunteer soccer players (29 women: age = 19.5 ± 1.8 years, height = 164 ± 9.1 cm, mass = 63.2 ± 7.1 kg; 15 men: age = 20.3 ± 2.9 years, height = 174 ± 6.7 cm, mass = 67.0 ± 9.5 kg) with at least 5 years of heading experience participated in the study. Potential participants were excluded from the study if they had a history of neurologic disorder, cervical spine injury, or head injury (eg, concussion) in the 6 months before data collection. Participants provided informed consent, and the institutional review boards at each school approved the study.

Instrumentation

Anthropometric Assessments. Height, mass, head-neck segment length, and neck girth were assessed for each participant. Body mass was measured in kilograms using a Kistler force plate (model 9287BA; Kistler Instrument Corp, Amherst, NY). Body mass was multiplied by the sex-specific head-neck segment to total body mass percentage (men = 8.26%, women = 8.20%) to determine head-neck segment mass.[16] Head-neck segment length was measured in centimeters with a metric tape measure from the seventh cervical vertebrae spinous process to the top of the head with the participant looking at an object at eye level. Neck girth was measured in centimeters with a metric tape measure as the circumference just above the thyroid cartilage. The intraclass correlation coefficient (ICC) (2,1) values for the anthropometric measurements were 0.99 (height), 0.99 (mass), 0.98 (head-neck segment length), and 0.99 (neck girth).

Isometric Strength Assessment. The microFet handheld dynamometer (Hoggan Health Industries, West Jordan, UT) was used to assess head-neck segment isometric flexor and extensor muscle strength with the participant seated and his or her torso stabilized. Flexor strength was assessed with the dynamometer placed in the center of the participant's forehead. Extensor strength was assessed with the dynamometer placed just above the participant's external occipital protuberance. The participant applied maximum force against the dynamometer for 3 seconds during each of 3 trials and rested for 30 seconds between trials. The peak values of each trial were averaged to determine the criterion measure. The ICC (2,1) value for this instrument was 0.96.

Head Kinematics Assessment (Resultant Linear Head Acceleration). A custom-fit mouthpiece (Figure 3) was fabricated for each participant by taking an impression of the teeth and making a 3-dimensional cast of the

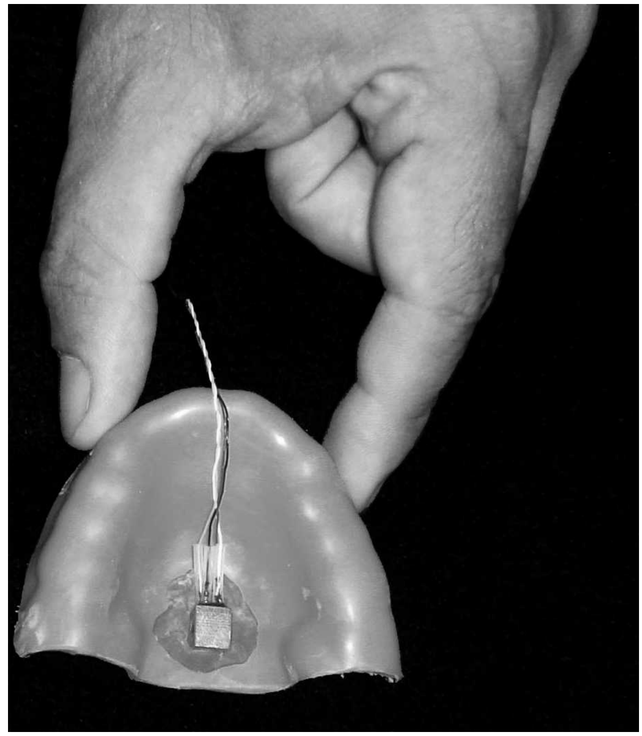

Figure 3. Mouthpiece accelerometer.

impression. Using a 12.7- × 12.7- × 4-cm sheet of thermoforming material (Henry Schein Inc, Melville, NY), we made the custom mouthpiece from the cast to cover the teeth and upper palate. On the second testing day, a triaxial accelerometer (model 35A; Endevco Corp, San Juan Capistrano, CA) was secured with dental wax to the upper palate portion of the mouthpiece.

The triaxial accelerometer was used to assess head acceleration data along the x-axis (anterior-posterior), y-axis (medial-lateral), and z-axis (superior-inferior). This 6- × 4- × 4-mm accelerometer has a mass of 1.1 g. Its range is ±1000 g (±5 V). Signals from the accelerometer were passed to a change and isotron signal conditioner (model 133; Endevco Corp), where they were filtered by a 4-pole low-pass Butterworth filter at 10 kHz and transmitted to the BNC connector system (Noraxon USA). This signal was converted to a digital signal using a 16-channel analog-to-digital converter card (KPCMCIA 12A1-C; Keithley Instruments Inc, Cleveland, OH). Data were stored in the MyoResearch software (version 2.11.1; Noraxon USA). The mouthpiece accelerometer has been reported to be a valid instrument for estimating head center-of-gravity acceleration.[17] The ICC (2,1) value of this instrument is 0.89.[15]

Electromyographic Assessment. We used the TeleMyo system to assess the preparatory and reactive electromyographic (EMG) activity of the left and right sternocleidomastoid and trapezius muscles. They were chosen because of their importance as superficial muscles that help to control head-neck flexion and extension and because of their use in previous head and neck research.[8,18] The skin over the right sternocleidomastoid and trapezius muscles was shaved, lightly abraded, and cleaned with 70% alcohol. Self-adhesive Ag/AgCl bipolar surface electrodes (Multi BioSensors Inc, El Paso, TX) with a diameter of 10 mm were placed on the skin 10 mm apart and parallel to the fiber orientation of the underlying muscle. The resistance between the paired electrodes was less than 2 kΩ and was verified with a standard digital multimeter (model 982017; Sears, Roebuck & Co, Hoffman Estates, IL). Placement of the electrodes was identified by palpating the midlength of the muscle's contractile component during an isometric contraction. A reference electrode was positioned on the skin over the right clavicle.

Signals from the muscle leads were passed to a battery-operated 8-channel FM transmitter worn by the participant. The signal was amplified (gain = 1000) with a single-ended amplifier (input impedance >10 MΩ) and filtered with a fourth-order Butterworth filter (10–500 Hz) and common mode rejection ratio of 130 dB at direct current (minimum 85 dB across the entire frequency of 10–500 Hz). An antenna receiver (Antenex Inc, Schaumburg, IL) with a sixth-order filter (gain = 2, total gain = 2000) further amplified the signal. The analog signal was converted to a digital signal by an analog-to-digital converter card and was stored in the MyoResearch software. The raw digital signal (for MVIC and trials) was sampled at a rate of 1000 Hz, rectified, and smoothed using a root mean square algorithm over a 20-millisecond moving window. All analyses were performed on processed EMG data during a 150-millisecond period before and a 250-millisecond period after force application. These periods were chosen because pilot data of muscle activity revealed the greatest activity during these periods around force application. Force application was determined using the FSR (diameter = 18 mm, force threshold = 1 N; Noraxon USA) positioned on the center of the forehead just below the hairline. The ICC (2,1) values for the EMG-dependent variables were 0.92 (peak activity) and 0.87 (muscle activity area).

We used the JUGS soccer machine (JUGS, Tualatin, OR), which is designed to duplicate any type of pass or shot. It has two 0.25-horsepower motors, and the speed can be adjusted through a dual dial on the interface. A ball can be fed through the machine, resulting in a shot with a maximum speed of 40 m/s (90 mph) and a maximum range of 73 m (239.5 ft). For this study, the initial velocity was 9.83 m/s (22 mph), the angle of projection was 40°, and the range was approximately 11 m (35 ft). Ball speed was comparable with that shown in previous research.[10,19]

Procedures

Familiarization Session. Volunteer participants met with the investigator, and the purpose and procedures of the study were explained. The participants provided informed consent and completed a physical activity health history questionnaire. Participants who met the inclusion criteria and had no exclusionary factors proceeded to the test session. Anthropometric measurements of neck length and girth were taken, and the mouthpiece was molded.

Test Session. Testing was conducted at a university laboratory. To begin testing, participants performed a neck warm-up consisting of 15 seconds of clockwise neck rotations, 15 seconds of counterclockwise neck rotations, and 2 repetitions of stretching for 15 seconds in flexion and 15 seconds in extension. For neck-strength assessment, participants performed 3 MVICs for neck flexion and extension. The FSR was secured to the forehead, and the custom-fit mouthpiece with the triaxial accelerometer that was affixed to the hard palate portion was placed in the participant's mouth. A 450-g ball inflated to 55 158 Pa was projected using the JUGS soccer machine. The participants aimed for a target in front of them, as if taking a shot on goal. Participants performed a total of 12 successful standing headers. A successful header was a header in which the ball made contact with the FSR, the participant verified appropriate contact, and the ball was directed at the target (Figure 4). Throughout testing, the investigator and the participant decided if the header was successful.

Data Analysis

Data were analyzed using descriptive and inferential statistics. Anthropometric and isometric strength variables were analyzed between the sexes using independent-samples *t* tests. Resultant head acceleration and HIC_{36} were analyzed using separate 2 (sex) × 3 (headgear) analyses of variance (ANOVAs) with repeated measures on the last factor. Head-neck segment muscle activity was analyzed using a 2 (sex) × 3 (headgear) multivariate analysis of variance (MANOVA) with repeated measures on the last factor. Pearson product moment correlations were performed between the anthropometric and strength measures and the resultant head accelerations. Post hoc *t* tests were used when appropriate. The α level was set at .05. We used SPSS for Windows (version 14.0; SPSS Inc, Chicago, IL) for data analysis.

RESULTS

Head-Neck Segment Anthropometric and Isometric Strength

The independent-samples *t* tests revealed sex differences (Table 1). Women exhibited 15% less head-neck segment mass, 5% less head-neck segment length, and 12% less neck girth compared with men. Women also exhibited 50% less isometric neck flexor strength and 53% less isometric extensor strength compared with men.

Linear Head Acceleration

The 2 (sex) × 3 (headgear) ANOVA revealed an interaction effect (Figure 5). Post hoc independent-samples *t* tests were performed for each headgear condition between the sexes. Women exhibited greater head accelerations versus men when wearing the Head Blast ($t_{1,42} = 2.89$, $P = .006$) and Full90 Select ($t_{1,42} = 3.79$, $P < .001$). With the control condition, we found no difference ($t_{1,42} = 1.42$, $P = .164$). Specifically, head acceleration in women was 32% greater than in men when wearing the Head Blast (21.52 ± 5.47 *g* versus 16.27 ± 6.16 *g*) and was 44% greater than in men when wearing the Full90 Select (21.84 ± 5.34 *g* versus 15.20 ± 5.83 *g*). Head acceleration was 10% greater in women than in men during the control condition (20.16 ± 4.12 *g* versus 18.25 ± 4.48 *g*).

Head Injury Criteria

The 2 (sex) × 3 (headgear) ANOVA revealed no interaction ($P = .069$, power = .532; Figure 6). Female HIC_{36} scores were 17.5 ± 7.5 during the control, 19.7 ± 9.5 during the Full90 Select, and 19.5 ± 10.2 during the Head

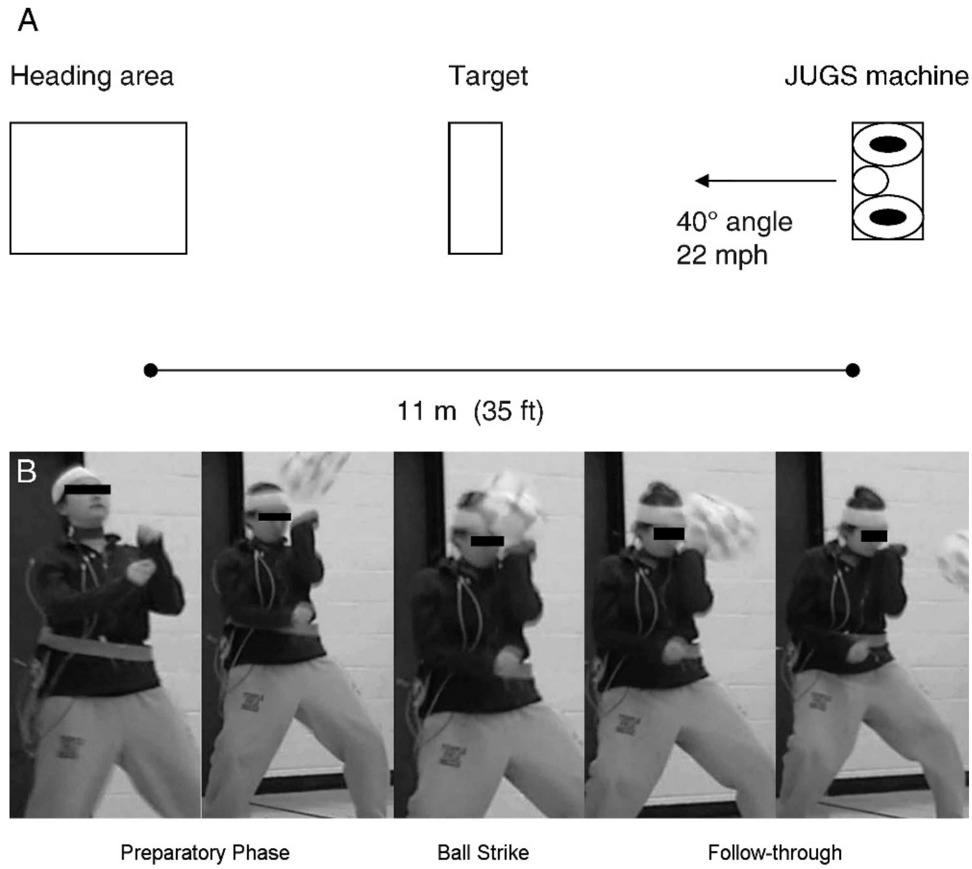

Figure 4. A, Schematic of test set-up. mph indicates miles per hour. B, Phases of soccer header.

Blast conditions. Male HIC_{36} scores were 15.9 ± 8.2 during the control, 13.9 ± 8.0 during the Full90 Select, and 16.0 ± 9.1 during the Head Blast conditions.

Head-Neck Segment Muscle Activity

The 2 (sex) × 3 (headgear) MANOVA revealed a main effect for sex ($F_{16,25} = 2.66$, $P = .014$) but not for headgear ($F_{32,9} = 2.55$, $P = .070$; Table 2) and no interaction effect ($F_{32,9} = 1.04$, $P = .501$). Follow-up individual ANOVAs for EMG peak and area did not indicate where significant differences existed.

Pearson Correlation Results

The Pearson correlations revealed relationships ($P < .05$) among the anthropometric variables, isometric strength, and resultant head accelerations (Table 3).

Table 1. Means (SDs) and Statistical Values for Anthropometric and Isometric Strength Data

Variable	Men (n = 15)	Women (n = 29)	$t_{1,42}$ Value	P Value
Head-neck segment mass, kg	6.14 (0.78)	5.19 (0.59)	−4.54	<.001
Head-neck segment length, cm	25.96 (1.45)	24.67 (1.30)	−3.01	.004
Neck girth, cm	37.08 (1.80)	32.60 (1.92)	−7.41	<.001
Isometric flexor strength, kg	15.88 (3.05)	8.00 (2.93)	−8.34	<.001
Isometric extensor strength, kg	21.18 (4.90)	10.02 (3.53)	−8.69	<.001

DISCUSSION

To our knowledge, this is the first study involving a reliable head impact model and a substantial number of human participants to examine the effect of sex and headgear on head kinematics. Using our head impact model, head acceleration and HIC_{36} data were much less than values associated with severe brain injury.[20–22] However, we were able to identify sex differences in head acceleration response associated with wearing soccer headgear. These differences indicate an increased concussion risk in women and a protective effect in men when

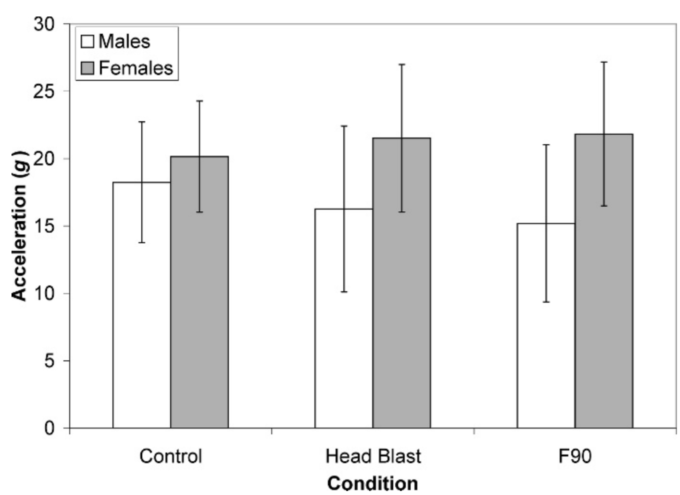

Figure 5. Resultant linear head acceleration in gravitational units. F90 indicates Full90 Select.

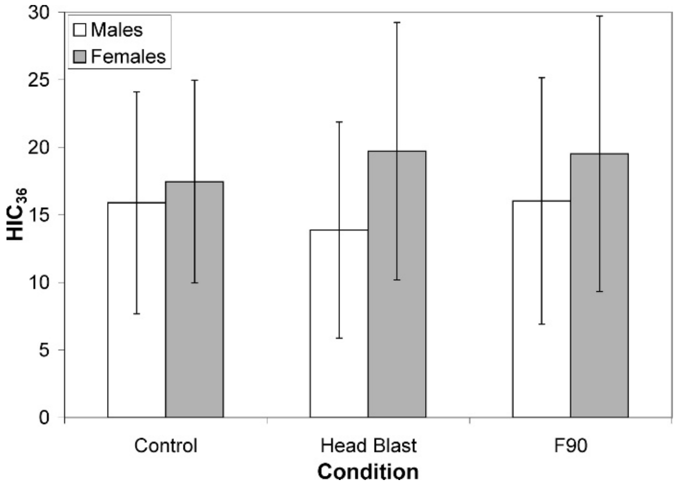

Figure 6. The Head Injury Criteria (HIC_{36}) across headgear conditions. F90 indicates Full90 Select.

wearing headgear. The degree to which these differences remain when using greater impact forces (eg, ball speeds) is unknown and warrants further study.

Researchers[4] recently reported that being female and not wearing headgear may increase the risk of sustaining a concussion in adolescent soccer players. Our data support the sex finding, because the women in our study experienced higher head accelerations during impact compared with men. However, headgear may not be beneficial for all athletes, because women in our study experienced the greatest head accelerations when wearing headgear. This finding may be a result of athletes feeling the need to strike the ball harder or of feeling safer when wearing the protective equipment.[23] This feeling causes athletes to attack balls more vigorously and adds to impact energy. This action had an apparent detrimental effect on women, possibly because of their lower head-neck segment stability qualities.

Consistent with previous research, women exhibited less effective head mass and neck strength compared with men.[7,8,14,16] Accordingly, their head impact acceleration during heading was 10% to 44% more than that of men. Sex differences in neck strength have been attributed to the amount of muscle tissue,[24,25] and in our study, women exhibited less (13%) neck girth compared with men. Enhancing neck muscle preactivation could have aided our participants, because neck muscle preactivation decreases head acceleration during force application.[6] However, consistent with previous research involving soccer players,[8] we found no sex differences in neck muscle activation strategies. Compared with men, women were at greater risk of concussion because they exhibited lower amounts of head stability qualities and did not increase neck muscle preactivation.

The headgear effectively reduced head acceleration for the men only. Previous research on soccer headgear has indicated that it is effective only when tested on a head form[9] during impacts with ball speeds of greater than 34 mph (15 m/s) or when tested during stiff impacts (eg, mounted to a rigid force plate).[12,14] Previous authors also have suggested that the thickness of headgear should be 15 mm[13] or more[10] to influence head impact acceleration. Thickness of the headgear in our study was less than these suggested values (Head Blast = 8 mm, Full90 Select = 11 mm), yet reductions in head acceleration were revealed for men. Participant head movement during the heading action combined with the ball speed of 22 mph (9.83 m/s) may have created impact energies similar to those reported in previous research that yielded effective headgear results.[9] The greater stability qualities also may have enabled men to create a stiffer impact, similar to previous headgear testing that yielded effective results.[9,12,14] Together these factors may have allowed the headgear to absorb impact energy and reduce head acceleration in men.

Limitations

Limitations of this study include the fact that we performed a straight-standing header only (versus rotational or jumping header), that we measured linear head acceleration (versus linear and rotational acceleration), and that we did not measure postimpact ball velocities. The straight-standing header was chosen to reduce variability during the headers so we could better assess a sex or headgear effect. We are the first to report the use of a mouthpiece accelerometer during a functional performance. This technique is an accurate predictor of the head center-of-gravity movement[17]; however, only linear acceleration can be assessed presently. Finally,

Table 2. Means (SDs) of Electromyographic Data

			Muscle Preactivity				Muscle Reactivity			
			Sternocleidomastoid		Trapezius		Sternocleidomastoid		Trapezius	
Sex	Headgear Condition	Measure[a,b]	Right	Left	Right	Left	Right	Left	Right	Left
Women	Control	Area	13.1 (16.2)	10.6 (11.3)	12.6 (16.0)	15.0 (22.8)	3.9 (2.6)	3.6 (2.8)	11.9 (16.3)	12.9 (17.4)
(n = 29)		Peak	67.6 (48.6)	54.6 (24.8)	72.8 (70.7)	80.4 (81.6)	42.9 (30.6)	42.6 (24.1)	89.0 (99.6)	92.5 (97.8)
	Head Blast	Area	14.2 (16.9)	11.1 (10.1)	9.5 (10.3)	11.6 (13.2)	5.1 (7.3)	3.3 (1.7)	9.2 (8.5)	11.0 (9.1)
		Peak	68.3 (67.3)	59.0 (38.4)	60.8 (44.1)	75.9 (53.3)	61.7 (97.5)	45.6 (31.9)	80.3 (75.6)	92.8 (98.6)
	Full90	Area	11.9 (15.9)	15.0 (21.1)	9.5 (12.2)	11.2 (13.9)	3.6 (4.0)	7.5 (15.8)	10.2 (10.6)	11.4 (13.7)
		Peak	62.9 (43.2)	102.9 (146.9)	73.5 (58.7)	84.6 (78.2)	34.8 (20.6)	83.1 (144.6)	86.5 (94.9)	98.0 (97.5)
Men	Control	Area	8.0 (8.4)	11.7 (22.2)	4.2 (2.9)	5.5 (3.6)	2.9 (2.3)	2.5 (2.9)	7.6 (7.9)	8.9 (5.1)
(n = 15)		Peak	70.9 (60.3)	84.5 (99.8)	47.1 (38.4)	60.7 (39.2)	41.9 (32.9)	36.9 (37.2)	71.1 (65.2)	79.4 (45.6)
	Head Blast	Area	7.4 (6.9)	8.1 (7.3)	4.5 (3.5)	6.1 (4.4)	2.2 (1.7)	2.7 (2.9)	6.6 (4.9)	11.1 (17.2)
		Peak	58.5 (44.9)	68.1 (55.8)	56.9 (49.4)	78.4 (72.8)	35.6 (31.3)	51.5 (61.3)	78.3 (70.0)	99.7 (96.2)
	Full90	Area	12.6 (17.1)	10.7 (11.2)	7.4 (11.9)	12.7 (18.4)	3.2 (2.9)	2.6 (1.7)	10.3 (13.3)	16.3 (18.6)
		Peak	62.6 (45.3)	58.5 (28.2)	58.3 (52.2)	83.7 (94.9)	31.6 (24.9)	34.7 (41.9)	66.4 (69.5)	99.2 (108.4)

[a] Area muscle activity, % × ms.
[b] Peak muscle activity, %.

Table 3. Pearson Correlation Coefficients (r) for Resultant Head Acceleration Versus Anthropometric and Strength Variables

	Resultant Head Acceleration		
Variable	Control	Head Blast	Full90 Select
Head-neck mass	$-.327^a$	$-.576^b$	$-.572^b$
Head-neck length	$-.237$	$-.353^a$	$-.360^a$
Neck girth	$-.184$	$-.423^b$	$-.484^b$
Neck flexor strength	$-.156$	$-.388^a$	$-.477^b$
Neck extensor strength	$-.101$	$-.316^a$	$-.443^b$

[a] Indicates $P \leq .05$.
[b] Indicates $P \leq .01$.

postimpact ball velocities would provide information regarding heading intensity with and without the headgear and, therefore, should be included in future research on soccer headgear.

CONCLUSIONS

We are the first to examine head kinematics during soccer heading with a substantial number of male and female participants. For our participant demographic, head acceleration was greater in women than in men when wearing the soccer headgear. Clinically, athletic trainers should be cautious when advising athletes or parents about the potential effects of the headgear. Based on our results, research is also needed to examine if soccer headgear may lead to increased head accelerations in children during soccer heading. This investigation is particularly important because children exhibit low amounts of head-neck segment stability qualities and headgear is primarily marketed toward this population.

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