Dental Traumatology

Dental Traumatology 2012; doi: 10.1111/j.1600-9657.2012.01135.x

Flexural properties and shock-absorbing capabilities of new face guard materials reinforced with fiberglass cloth

Keisuke Abe[1], Hidekazu Takahashi[2], Hiroshi Churei[1], Naohiko Iwasaki[2], Toshiaki Ueno[1]

[1]Sports Medicine/Dentistry, Graduate School of Medical and Dental Sciences, Tokyo Medical and Dental University; [2]Oral Biomaterials Engineering, Course of Oral Health Engineering, Faculty of Dentistry, Tokyo Medical and Dental University, Tokyo, Japan

Key words: face guard; fiberglass cloth; fiber-reinforced plastic; shock absorption

Correspondence to: Hiroshi Churei, Sports Medicine/Dentistry, Graduate School of Medical and Dental Sciences, Tokyo Medical and Dental University, Yushima 1-5-45, Bunkyo-ku, Tokyo, 113-8549, Japan
Tel.: +81 3 5803 5867
Fax: +81 3 5803 5867
e-mail: chu.spmd@tmd.ac.jp

Accepted 3 March, 2012

Abstract – *Aim*: Experimental materials incorporating fiberglass cloth were used to develop a thin and lightweight face guard (FG). This study aims to evaluate the effect of fiberglass reinforcement on the flexural and shock absorption properties compared with conventional thermoplastic materials. *Material and Method*: Four commercial 3.2-mm and 1.6-mm medical splint materials (Aquaplast, Polyform, Co-polymer, and Erkodur) and two experimental materials were examined for use in FGs. The experimental materials were prepared by embedding two or four sheets of a plain woven fiberglass cloth on both surfaces of 1.5-mm Aquaplast. The flexural strength and flexural modulus were determined using a three-point bending test. The shock absorption properties were evaluated for a 5200-N impact load using the first peak intensity with a load cell system and the maximum stress with a film sensor system. *Results and Conclusions*: The flexural strength (74.6 MPa) and flexural modulus (6.3 GPa) of the experimental material with four sheets were significantly greater than those of the 3.2-mm commercial specimens, except for the flexural strength of one product. The first peak intensity (515 N) and maximum stress (2.2 MPa) of the experimental material with four sheets were significantly lower than those of the commercial 3.2-mm specimens, except for one product for each property. These results suggest that the thickness and weight of the FG can be reduced using the experimental fiber-reinforced material.

Maxillofacial traumatic injuries sometimes occur in contact sports such as soccer, rugby football, and American football (1–5). The face guard (FG) has been introduced as a protective device for the early and safe return to play of athletes after sustaining maxillofacial traumatic injury, and the effectiveness of FGs has been widely recognized (6–16). However, the guidelines for FG fabrication have not yet been established, and FGs have been fabricated through trial and error with materials and designs selected based on clinical usage (17).

A FG is required to fulfill the following three requirements: (i) to protect the player from re-injury (protection ability), (ii) to not hurt other players (safety), and (iii) to not narrow the player's field of vision (maintain the performance) (9). In particular, the first two requirements are stated in Article 4 of the Laws of the Game for the Federation lnternationale de Football Association (FIFA) as follows: 'a player may use equipment other than the basic equipment, provided that the sole purpose is to provide physical protection and that no danger is posed to the wearer or any other players' (18). It has already been reported that any effects on the field of vision must be minimized to maintain the performance (requirement 3), which is illustrated by the clinical effectiveness of the FG design based on objective data from visual field tests (19, 20).

According to a questionnaire given to the players after FG usage, they were satisfied with the protective ability of the FG but were dissatisfied with its fit, claiming that it slipped off while playing and narrowed their field of vision; therefore, thinner and lighter FGs are required. In particular, professional players insisted on improvements to the FG's field of vision and bulkiness (16, 20).

Typically, thermoplastic resins have been used as the FG material. The molding temperature of the thermoplastic resin is a very important aspect of the FG fabrication process. If the required molding temperature is low enough, then expensive vacuum and/or pressure thermoforming machines are not required because the materials can be easily molded using hot water and finger pressure (12, 15–17). However, thermoplastic resins with low molding temperatures exhibit relatively low mechanical properties; therefore, FGs that use these thermoplastic resins are generally thicker than those that use high molding temperature thermoplastic resins (21).

According to previous research on the impact absorption properties of FGs and mouth guards (22–25), a combination of hard thermoplastic material and soft cushioning material can provide remarkable shock absorption properties. To further improve the capacity for shock absorption, lining the inner surface of the hard thermoplastic material with the cushioning material is more effective than placing the cushioning material on the outer surface of the hard thermoplastic material. However, the outer surface of the FG then has to be covered with soft material to prevent injury to the wearer and other players (15–18, 22). As a result, a thick FG consisting of three layers narrowed the field of vision for the wearing player (20). To solve this problem, a reduction in the thickness of the hard thermoplastic material was attempted by reinforcing the conventional thermoplastic material with fiberglass. Thermoplastic materials that are reinforced with fiberglass are widely known as fiber-reinforced plastic (FRP) materials. This study aimed to evaluate the effect of fiberglass reinforcement on the flexural properties and shock absorption capability compared with conventional thermoplastic materials in an effort to develop a thin and lightweight FG. The null hypothesis was that the flexural properties and shock absorption capability of the 1.6-mm-thick experimental FRP FG materials are not significantly different from those of the 3.2-mm-thick commercial FG materials.

Materials and methods

Four commercial thermoplastic resins for medical splints, including Aquaplast (AP; Homecraft Rolyan, Huthwaite, North Nottingham, UK), Polyform (PF; Homecraft Rolyan), Co-polymer (CP; Advanfit, Kumamoto, Japan), and Erkodur (ED; Erkodent, Pfalzgrafenweiler, Germany), as well as two experimental FRP materials, were examined. Thicknesses of 1.6 mm (1/16 inch) and 3.2 mm (1/8 inch) were selected for the commercial thermoplastic resins (Table 1). For CP and ED, no 1.6-mm-thick products were available; therefore, 1.6-mm-thick specimens were prepared from 3.2-mm-thick product using a hot press machine (modified AH-1T; As One Co., Osaka, Japan) at 150°C for ED and 190°C for CP with a final compression load of 5000 N.

The softening temperatures of the commercial thermoplastic resins were confirmed using the needle penetration method. A 15 mm × 15 mm piece of each 1.6-mm-thick commercial thermoplastic resin was placed on a stainless steel plate and heated in a hot air drying oven (INCUBATOR CI-610; Advantec Co., Chiba, Japan) at heating intervals of 5°C for a storage period of 15 min. A penetration needle device (with a 2.0-mm-diameter needle weighing 50 g) was placed perpendicular to the surface of the thermoplastic resin. The softening temperature of the material was determined as the temperature at which the needle penetrated 1.0–1.5 mm into the resin. Five specimens of each material were evaluated.

Two experimental FRP materials were prepared from AP and sheets of plain woven E-fiberglass cloth (M100X 104H; Unitika, Osaka, Japan; density: 100 g m^{-2}) using the heat press method: one containing two sheets of fiberglass cloth with one sheet on each surface (AP-G2) and the other containing four sheets of fiberglass cloth with two sheets on each surface (AP-G4). A 3.2-mm-thick AP sheet was prepared to a 250 mm length and 150 mm width using an ultrasonic cutter (Labo Sonic Cutter model NE87; Nakanishi Inc., Tochigi, Japan). A piece of AP was powdered by frost shattering using a vibration mixer mill (MM301; Retsch GmbH, Haan near Düsseldorf, Germany) and liquid nitrogen. The sheets of fiberglass cloth were cut to 300 mm in length and 200 mm in width. The AP sheet was placed between the fiberglass cloth sheets and pressed using a hot press machine at 180°C with a final compression load of 6000 N. Additionally, powdered AP was sprinkled on the area where the thermoplastic material poorly impregnated the fiberglass cloth, and the material was hot pressed again at a final compression load of 8000 N to make it 1.5 mm in thickness. Therefore, the fiberglass cloth was not exposed on the surfaces of either AP-G2 or AP-G4.

Specimens for the three-point bending test (100 mm in length and 15 mm in width) were prepared using an ultrasonic cutter and waterproof abrasive paper. After measuring the specimens with a micrometer (293-421-20; Mitsutoyo, Kanagawa, Japan; minimum reading: 0.001 mm), the three-point bending test was conducted with a support span width of 40 mm and a crosshead speed of 1.0 mm min^{-1} using a universal test machine (Model 1123; Instron, Canton, MA, USA). The data from the three-point bending test were stored on a personal computer. The flexural strength, flexural modulus, and bending load at a 1-mm middle-span deflection per 1 mm specimen width (BL1) were calculated based on the following equations using statistical software (Series IX; Instron):

$$\text{Flexural strength} = 3Fl/2bh^2$$
$$\text{Flexural modulus} = F_1 l^3/4bh^3 d$$
$$\text{BL1} = F_2/b$$

where F is the maximum load (N), l is the width of the support span (mm), b is the width (mm) of the specimen, h is the height (mm) of the specimen, F_1 is the load (N) at a point in the straight-line portion of the trace, d is the deflection (mm) at load F_1, and F_2 is the load (N) at the 1-mm middle-span deflection. Five specimens of each material were examined.

Specimens for the shock absorption test (100 mm in length and 100 mm in width) were prepared using the ultrasonic cutter and were covered with a sheet of cushioning material (1.6-mm-thick Neoprne; Homecraft Rolyan) on both sides using a cyanoacrylate adhesive. The impact load of a freefalling object (500-g steel rod with a 3/16-inch-diameter rounded end) from a height of 25 cm was applied to the specimens using an impact testing machine (modified IM-201; Tester Sangyo Co., Saitama, Japan) (Fig. 1).

Two measuring systems were used: one was a load cell system, and the other was a film sensor system. For the load cell system, three dynamic compression load cells (LMB-A-2KN; Kyowa Electronic Instruments Co., Tokyo, Japan) were placed 120° apart below a 10-mm-thick stainless steel platform, and the specimen was placed at the center of the platform. Changes in load during and after the applied impact loading (with and without specimens) were recorded via a sensor interface (EDX-100A; Kyowa Electronic Instruments Co.) to a personal computer using an analysis software (DCS-100A; Kyowa Electronic Instruments Co.) at a

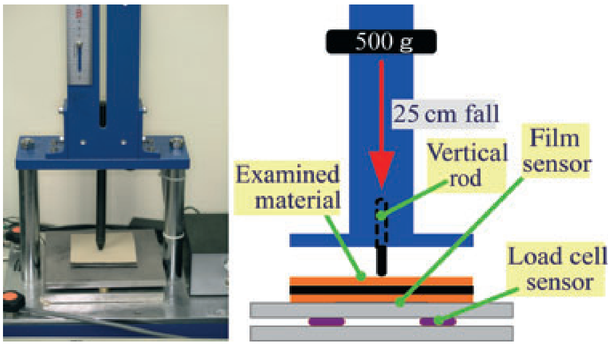

Fig. 1. Shock absorption test. The experimental force of impact was applied by a free-fall drop and impact testing machine (modified IM-201; Tester Sangyo Co., Saitama, Japan) that consisted of a free-falling object (500 g) and a vertical rod. Shock absorptions were comparatively assessed under the two combinations of examination materials.

sampling rate of 20 kHz. The sum of the loads recorded by the three load cells was calculated, and the maximum sum of the loads after the applied impact load was registered as the first peak intensity. The results without the specimen were recorded as a reference.

For the film sensor system, a precut type of film sensor (Presheet: Prescale sheet type; Fujifilm Business Supply Co., Tokyo, Japan) was placed under the specimen. There were seven measurable pressure ranges of film sensors; an appropriate sheet was selected for each condition. The induced red patch at the load impact point (with and without the specimens) was recorded using a camera system (Data Shot FPD-100; Fujifilm Business Supply Co.). The maximum stress and the impressed area were analyzed by analysis software (Data Shot FPD-100S; Fujifilm Business Supply Co.). The results without the specimen were recorded as a reference. Five impact loads were applied for each specimen, and five specimens were examined for each condition. The obtained results were analyzed using one-way analysis of variance with Tukey's honestly significant difference multiple comparison test; the analysis was performed using statistical software (JMP ver. 9; SAS Institute Inc., Cary, NC, USA) with a significance level of 5%.

Results

The measured softening temperatures were consistent among the five specimens for each material. The softening temperatures ranked as follows: AP (70°C), PF (70°C), ED (150°C), and CP (190 °C).

Except for AP-G2 and AP-G4, none of the specimens fractured during the three-point bending test; therefore, the maximum stress during the test was considered the flexural strength. The flexural strengths of the 1.6-mm specimens ranged from 18.5 to 67.2 MPa, while those of the 3.2-mm specimens ranged from 32.1 to 86.3 MPa. The flexural strengths of the 3.2-mm specimens were greater than those of the 1.6-mm specimens for the same material. The flexural strength of AP-G4 was greater than those of the 3.2-mm specimens, except for ED-32.

The flexural moduli of the commercial medical splint materials ranged from 0.5 to 2.3 GPa but were fairly consistent for the same material, regardless of thickness. The flexural moduli of the experimental FRP materials were greater than that of AP; the flexural modulus of AP-G4 (6.3 GPa) was significantly greater than those of the other materials.

The BL1s of the 1.6-mm specimens ranged from 0.12 to 0.56 N mm^{-1}, while those of the 3.2-mm specimens ranged from 0.97 to 3.33 N mm^{-1}. The BL1s of the 3.2-mm specimens were greater than those of the 1.6-mm specimens when the same material was used. The BL1 of AP-G4 (1.35 N mm^{-1}) was significantly greater than those of the other specimens, except for PF-32, CP-32, and ED-32.

After the shock absorption test, an obvious trace of the steel rod was observed on each specimen, and extensive radial cracks from the trace were found on all of the ED-16 specimens. The first peak intensity without the specimen was 5256.5 ± 343.6 N, and the first peak intensity decreased when the specimens were inserted. The first peak intensities of the 1.6-mm commercial medical splint specimens ranged from 923 to 1221 N and were significantly greater than those of the 3.2-mm specimens, which ranged from 415 to 571 N. The first peak intensity of AP-G4 (515 N) was significantly lower than those of the other 1.6-mm specimens but not significantly different from those of the 3.2-mm specimens, except for PF-32.

The maximum stress of ED-32 could not be determined because it was below the stress range of the film sensor sheet. The maximum stresses of the 1.6-mm commercial medical splint specimens ranged from 25.6 to 45.7 MPa and were significantly greater than those of the 3.2-mm specimens, which ranged from below 0.5 to 5.2 MPa. The maximum stress of AP-G4 (2.2 MPa) was significantly lower than those of the 1.6-mm commercial medical splint specimens and not significantly different from those of the 3.2-mm specimens, except for ED-32.

The impressed stress distribution areas of the 1.6-mm commercial medical splint specimens ranged from 34.8 to 47.5 mm^2 and were significantly greater than those of the 3.2-mm specimens, which ranged from 0.0 to 29.2 mm^2. The impressed stress distribution area of AP-G4 (4.9 mm^2) was significantly smaller than those of the other 1.6-mm specimens and not significantly different from those of the 3.2-mm specimens, except for ED-32 (Table 2).

Discussion

The 3.2-mm-thick thermoplastic resins for medical splints that were examined in the present study were used as FG materials in previous studies (12, 15–17, 19). However, the available information of these thermoplastic resins' properties was limited. Although minimum properties have not yet been established, FGs made with a 3.2-mm thermoplastic resin have been successful in clinical use. Therefore, the minimum requirements for FG materials are considered to be the weakest values of the 3.2-mm thermoplastic resins.

The measurement of the softening temperature was important to understanding the moldable temperature

(12, 17). The needle penetration test used in the present study was easy to perform and confirmed the softening temperatures. According to the softening temperature results, AP and PF were confirmed to be formable using hot water and finger pressure. Therefore, AP was selected for the experimental FRP.

There are two types of fiberglass: short-rod fiberglass and continuous fiberglass. Continuous fiberglass is considered more effective at improving the mechanical properties of the FRP material; therefore, fiberglass cloths are commonly used for FRP. There are several weave types for fiberglass cloths, such as plain, satin, leno, and twill. Moreover, the density, thickness, and mesh size of the fiberglass cloth influence the reinforcement ability. Plain woven fiberglass cloth was selected based on the penetration ability of thermoplastic resin during pilot studies. The location of the fiberglass was also an important factor for improving the mechanical properties of the FRP. Embedding fiberglass on both surfaces improved the toughness from 0.1 to 3.8 MPa (26). Therefore, the fiberglass cloth was embedded on both surfaces.

The three-point bending test was used to determine the flexural strength, flexural modulus, and BL1. The flexural strength is the maximum stress achieved during the test, which indicates how much stress could be applied before fracture. The flexural modulus is the material constant of deflection, which represents how easily the material can be bent within the elastic deformation limits. The BL1 is also an indicator of how easily the material can be bent and is defined as the required load for a 1-mm-wide specimen to deflect 1 mm (including both elastic and plastic deformation). The 1-mm deflection was selected because a common cushioning material (1.6-mm-thick neoprne rubber) for FGs could easily compressed to 0.6 mm without a heavy load (12, 15–17, 22, 23). The flexural strengths, flexural moduli, and BL1s of the 3.2-mm medical splint materials showed similar values to those that were previously reported (21). A constant cross-head speed was used for the three-point bending test. The actual thicknesses of the 3.2-mm medical splint materials were around 3.0 mm; therefore, the strain increasing ratio of the 3.2-mm specimens was 6.6 (which is the cube of 3.0/1.6) times greater than that of the 1.6-mm specimens. A greater strain increase ratio generally increases the flexural strength and modulus of a material with visco-elastic properties. Consequently, the flexural strength and flexural modulus of the 1.6-mm specimens were lower than those of the 3.2-mm specimens; therefore, the BL1s of the 3.2-mm specimens were expected to be 6.6 times greater than those of the 1.6-mm specimens. The 1.6-mm specimens might also show greater plastic deformation than the 3.2-mm specimens. As a result, the BL1s of the 3.2-mm medical splint materials were between five and eight times greater than those of the 1.6-mm medical splint materials.

The experimental FRP material showed greater flexural strength, flexural modulus, and BL1, and AP-G4 was better than AP-G2. The minimum required flexural strength and BL1 were considered to be those of AP-32 (12, 15–17, 22). Taking these values into account, CP-16, ED-16, AP-G2, and AP-G4 were determined to be sufficient in terms of flexural strength, but only AP-G4 was sufficient in terms of BL1.

The load cell system used in the present study could monitor the load transmitted below the FG materials over time; therefore, the intensity of and change in the impact load were recorded. The impact force required to fracture the human maxillofacial bone has been reported to be between 4930 and 5780 N (27); therefore, the impact load of a free-falling object (500 g) from a 25-cm height (5256 N) was used in the present study. The impact load absorption capability, which is the ratio of the decreased impact load by the FG material to the original impact load, has often been discussed (22, 23, 28). Previous research using an impact load system similar to that of the present study reported that the impact load absorption capabilities of the medical splint materials ranged from 85 to 88% (22). These results agree with the results of the present study (89–92%). There were no obvious relationships between the impact load absorption capability and the flexural properties (flexural strength, flexural modulus, and BL1).

The first peak intensities of the experimental FRP (AP-G2 and AP-G4) were lower than those of the 1.6-mm medical splint materials, and that of AP-G4 was almost the same as those of the 3.2-mm medical splint materials, except for PF-32. Reinforcement with fiberglass cloth increased the flexural modulus of the

Table 1. Materials used in the present study

Type	Product name	Manufacture	Composition	Lot (thickness)	Code
Commercial medical splint	Rolyan Aquaplast Watercolors, Solid Charcoal	Homecraft Rolyan	Polycaprolactone	01003C12 (1.6 mm)	AP-16
				09608K02 (3.2 mm)	AP-32
	Rolyan Polyform, Solid. White	Homecraft Rolyan	Polycaprolactone	NA (1.6 mm)	PF-16
				NA (3.2 mm)	PF-32
	Co-polymer sheet, Yellow	Advanfit	Polypropylene polyethylene CP	NA (1.6 mm)[1]	CP-16
				NA (3.2 mm)	CP-32
	Erkodur, Clear	Erkodent	Polyethylene terephthalate	NA (1.6 mm)[1]	ED-16
				10607 (3.2 mm)	ED-32
Experimental material			AP + fiberglass cloth (2 sheets)	(1.5 mm)	AP-G2
			AP + fiberglass cloth (4 sheets)	(1.5 mm)	AP-G4

AP, Aquaplast; CP, co-polymer; ED, Erkodur; NA, not available; PF, polyform.
[1]Prepared using 3.2-mm specimen.

Table 2. Results of the three-point bending test and shock absorption test

Properties	AP-16	PF-16	CP-16	ED-16	AP-G2	AP-G4	AP-32	PF-32	CP-32	ED-32
Three-point bending test										
Flexural strength (MPa)	18.5 ± 1.3^{F}	20.5 ± 1.0^{F}	33.8 ± 1.2^{E}	67.2 ± 2.1^{C}	41.6 ± 1.2^{D}	74.6 ± 6.1^{B}	32.1 ± 1.4^{E}	35.4 ± 1.8^{E}	45.3 ± 0.9^{D}	86.3 ± 0.4^{A}
Flexural modulus (MPa)	468 ± 26^{F}	$874 \pm 43^{E, F}$	$1418 \pm 97^{D, E}$	$2061 \pm 76^{B, C, D}$	2596 ± 136^{B}	6302 ± 995^{A}	567 ± 33^{F}	$1439 \pm 106^{D, E}$	$1577 \pm 88^{C, D}$	$2250 \pm 12^{B, C}$
BL1 (N mm^{-1})	0.12 ± 0.01^{H}	$0.23 \pm 0.01^{G, H}$	$0.38 \pm 0.02^{G, F}$	0.56 ± 0.03^{F}	0.56 ± 0.04^{F}	1.35 ± 0.22^{D}	0.97 ± 0.06^{E}	2.21 ± 0.16^{C}	2.58 ± 0.08^{B}	3.33 ± 0.02^{A}
Shock absorption test										
First peak intensity (N)	1014 ± 19^{C}	923 ± 48^{D}	1152 ± 30^{B}	1221 ± 106^{A}	913 ± 33^{D}	515 ± 71^{F}	571 ± 18^{E}	415 ± 16^{G}	555 ± 20^{E}	569 ± 20^{E}
Maximum stress (MPa)	30.5 ± 2.6^{C}	25.6 ± 5.7^{C}	37.4 ± 1.9^{B}	45.7 ± 5.7^{A}	25.0 ± 1.3^{C}	2.2 ± 0.5^{D}	5.2 ± 0.9^{D}	1.4 ± 0.2^{D}	1.1 ± 0.1^{D}	Below 0.5
Stress impressed area (mm^{2})	40.7 ± 1.9^{B}	39.1 ± 3.4^{B}	$34.8 \pm 2.7^{B, C}$	47.5 ± 3.7^{A}	30.0 ± 4.2^{C}	4.9 ± 2.4^{D}	29.2 ± 4.8^{C}	2.6 ± 2.1^{D}	0.1 ± 0.1^{D}	ND

AP, Aquaplast; BL1, bending load at 1-mm middle-span deflection per 1 mm width; CP, co-polymer; ED, Erkodur; PF, polyform; ND, not detected.
Values with the same superscript letter were not significantly different ($P < 0.05$).

experimental FRP, which suggests that the stress would be widely distributed, causing the cushioning material to compress beneath the FRP (14). As a result, the first peak intensities of the experimental FRP showed lower values.

The film sensor sheet can precisely record the impact area by changing the color to red (29). From this discoloration, the level of stress can be analyzed using a camera or scanner and special software. Each film sensor sheet has a limited range of sensitivity; therefore, four types of film sensor sheets were used in the present study. However, the impact area below 0.5 MPa and over 130 MPa could not be detected. As a result, the maximum stresses were measured for all except for the reference and ED-32 tests. Generally, the specimen with a lower maximum stress showed a smaller impressed area. The FG material with a lower maximum stress is preferable for protecting the injured area. Therefore, the 3.2-mm medical splint was more suitable than the 1.6-mm medical splint. A decrease in the maximum stress meant a good absorption of the impact load and an even distribution of the impact load. As a result, the maximum stress of ED-32 could not be detected because it was under the threshold value of the film sensor. The maximum stress and the impressed area of AP-G2 and AP-G4 were lower than those of the 1.6-mm medical splints, and AP-G4 showed a lower maximum stress and smaller impressed area than AP-32. These results suggest that AP-G4 possesses suitable shock absorption properties for FGs despite being relatively thin.

In the present study, two or four sheets of fiberglass cloth were embedded in thermoplastic resin that could be softened by hot water. The obtained results suggest that reinforcement by fiberglass cloth is effective at increasing the flexural strength, flexural modulus, and BL1 and decreasing the first peak intensity, maximum stress, and impressed area. AP-G4 showed better characteristics than AP-32. If more sheets are embedded, the characteristics might be even better; however, the impregnation of thermoplastic resin into the fiberglass cloth might be more difficult. Therefore, a new technique and a more easily moldable thermoplastic resin are necessary to manufacture a FRP material with a large number of fiberglass cloth sheets. If a medical splint material with a higher softening temperature (CP or ED) were used for the experimental FRP material, we would expect the mechanical and shock absorption properties of the material to be better; however, the handling properties would be worse, and special equipment would be required for manufacturing (6–9, 11, 13, 14). Thus, the null hypothesis, in which the flexural and shock absorption properties of the 1.6-mm-thick experimental FRP FG materials are not significantly different from those of the 3.2-mm-thick commercial FG materials, was partially accepted. The results of the present *in vitro* study were obtained using a simple plane-shaped specimen. Therefore, the evaluation of FG materials using more complicated face-shaped specimens should be conducted in the future. Moreover, the minimum property requirements for the FG materials have not been established. Further research is needed to confirm these properties.

The experimental material examined in the present study was expected to reduce the thickness and weight of the FG. Therefore, the clinical assessment of the thin and lightweight FG using this experimental material is necessary to determine whether players are satisfied with the improvements in the field of vision and bulkiness of this FG.

Conclusions

Within the limitations of this *in vitro* investigation, the 1.5-mm-thick experimental FRP material with four embedded sheets of fiberglass cloth showed better or similar mechanical properties and shock absorption capability than the 3.2-mm-thick medical splint materials. These results suggest that the thickness and weight of the FG could be reduced using the experimental FRP material.

Acknowledgements

The authors would like to express their sincere gratitude to Prof. Masahiro Kobayashi at the Chiba Institute of Technology for his invaluable suggestions and assistance throughout this research. This work was supported by the Kozuki Foundation for sports and education.

References

1. Maladie're E, Bado F, Meningaud JP, Guilbert F, Bertrand JC. Aetiology and incidence of facial fractures sustained during sports: a prospective study of 140 patients. Int J Oral Maxillofac Surg 2001;30:291–5.
2. Delilbasi C, Yamazawa M, Nomura K, Iida S, Kogo M. Maxillofacial fractures sustained during sports played with a ball. Oral Surg Oral Med Oral Pathol Oral Radiol Endod 2004;97:23–7.
3. Exadaktylos AK, Eggensperger NM, Eggli S, Smolka KM, Zimmermann H, Iizuka T. Sports related maxillofacial injuries: the first maxillofacial trauma database in Switzerland. Br J Sports Med 2004;38:750–3.
4. Mourouzis C, Koumoura F. Sports-related maxillofacial fractures: a retrospective study of 125 patients. Int J Oral Maxillofac Surg 2005;34:635–8.
5. Fuller CW, Junge A, Dvorak J. A six year prospective study of the incidence and causes of head and neck injuries in international football. Br J Sports Med 2005;39:13–9.
6. Kaplan S, Driscoll CF, Singer MT. Fabrication of a facial shield to prevent facial injuries during sporting events: a clinical report. J Prosthet Dent 2000;84:387–9.
7. Heise M, Eufinger H, Rarreck T. Individual facial protection masks after fracture treatment of the nasal bone and the zygomatic arch in professional soccer players (German, English abstract). Mund Kiefer Gesichtschir 2001;5:320–2.
8. Oriya S, Shiraishi M. Treatment of the professional soccer player who returned quickly to play after a mandibular fracture (Japanese, English abstract). J Sports Dent 2001;4:28–31.
9. Nishino S, lrie Y, Nakajima H, Masunari M, Amano M, Kawamura K. Sports and protectors Facial guard for soccer players-Fabrication of nasal bone protectors (Japanese). J Clin Sports Med 2004;21:433–40.
10. de Baat C, Peters R, van lperen-Keiman CM, de Vleeschouwer M. A facial shield for a sportsman with an orofacial trauma (Dutch, English abstract). Ned Tijdschr Tandheelkd 2005;112: 177–80.
11. Yoshida K, Nishiyama S, Ito M. Fabrication of a facial guard after zygomatic arch fracture (Japanese). PO Academy J 2006;14:150–1.
12. Morita R, Shimada K, Kawakami S. Facial protection masks after fracture treatment of the nasal bone to prevent re-injury in contact sports. J Craniofac Surg 2007;18:143–5.
13. Cascone P, Petrucci B, Ramieri V, Marianetti TM. Security hi-tech individual extra-light device mask: a new protection for [soccer] players. J Craniofac Surg 2008;19:772–6.
14. Procacci P, Ferrari F, Bettini G, Bissolotti G, Trevisiol L, Nocini PF. Soccer-related facial fractures: postoperative management with facial protective shields. J Craniofac Surg 2009;20:15–20.
15. Fujino S, Churei H, Abe K, Miura H, Takahashi T, Ueno T. A custom faceguard for a soccer player injured with a complex fracture of the zygomatic bone and orbit: a case report (Japanese, English abstract). J Sports Dent 2010;13:108–13.
16. Churei H, Abe K, Fujino S, Sharika S, Chowdhury RU, Saito S et al. Clinical effectiveness of a custom faceguard for a futsal player injured with a nasal bone fracture for early and safe return: a case report. Int J Sport Dent 2011;4:34–40.
17. Ueno T, Churei H. Fabrication technique for custom faceguard with thermoforming material. Int J Sports Dent 2008;1:67–71.
18. Federation lnternationale de Football Association (2011). Laws of the game 2011/2012. http://www.fifa.com/mm/document/affederation/generic/81/42/36/lawsofthegame_2011_12e.pdf [accessed on 12 December 2011].
19. Churei H, Fujino S, Ihara C, Sumita Y, Ueno T. Clinical effectiveness of a custom faceguard for a rugby football player injured with a nasal bone fracture – feedback from visual field test (Japanese). Jpn J Clin Sports Med 2007;15:163.
20. Ueno T, Churei H, Abe K, Fujino S, Takahashi T. Clinical assessment of custom faceguards provided for sport-related maxillofacial bone fracture cases (Japanese). J Sports Dent 2011;14:108.
21. Abe K, Churei H, Takahashi H, Ueno T. Flexural properties of a faceguard core material measured by three-point bending test (Japanese, English abstract). J Sports Dent 2011;15:17–22.
22. Churei H, Yokota K, Takahashi H, Ueno T. Evaluation of fundamental physical-properties of face guard materials -1. Impact absorption (Japanese, English abstract). J Sports Dent 2008;11:90–5.
23. Churei H, Abe K, Yokota K, Takahashi H, Ueno T. Fundamental evaluation of the physical properties of face guard materials-II. Impact absorption (effect of perforation pattern) (Japanese, English abstract). J Sports Dent 2010;14:17–21.
24. deWet FA, Heyns M, Pretorius J. Shock absorption potential of different mouth guard materials. J Prosthet Dent 1999;82: 301–6.
25. Patrick DG, Van Noort R, Found MS. Scale of protection and the various types of sports mouthguard. Br J Sports Med 2005;39:278–81.
26. Dyer SR, Lassila LV, Jokinen M, Vallittu PK. Effect of cross-sectional design on the modulus of elasticity and toughness of fiber-reinforced composite materials. J Prosthet Dent 2005;94:219–26.
27. Nahum AM, Gatts JD, Gadd CW, Danforth J. Impact Tolerance of the Skull and Face. Proceedings of the 12th Stapp Car Crash Conference, New York; 1968. October 22–23, p. 302–16.
28. Takeda T, Ishigami K, Kawamura S, Nakajima K, Shimada A, Regner CW. The influence of impact object characteristics on impact force and force absorption by mouthguard material. Dent Traumatol 2004;20:12–20.
29. Phunthikaphadr T, Takahashi H, Arksornnukit M. Pressure transmission and distribution under impact load using artificial denture teeth made of different materials. J Prosthet Dent 2009;102:319–27.