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Characterization of Water-Clear Polymeric Gels for Use as Radiotherapy Bolus

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Abstract

Our purpose was to investigate polymeric gels for use as a highly transparent radiotherapy bolus and determine the relevant physical and dosimetric properties. We first quantified tensile properties (maximum stress, strain, and Young modulus) for various polymeric gels, along with a commercial bolus product in order to illustrate the wide variety of potential materials. For a select polymeric gel with tensile properties similar to currently used radiotherapy bolus, we also evaluated mass and electron density, effective atomic number, optical transparency, and percent depth dose in clinical megavoltage photon and electron beams. For this polymeric gel, mass density was 872 $\pm$ 12 and 896 $\pm$ 13 g/cm^3 when measured via weight/volume and computed tomography Hounsfield units, respectively. Electron density was 2.95 $\pm$ 0.04 $\times 10^{23}$ electrons/cm^3. Adding fused silica (9% by weight) increases density to that of water. The ratio of the effective atomic number to that of water without and with added silica was 0.780 and 0.835 at 1 MeV, 0.767 and 0.826 at 6 MeV, and 0.746 and 0.809 at 20 MeV. Percent depth dose for 6 MV photons was within 2% of water within the first 2.5 cm and after scaling by the density coincided within 1% out to >7 cm. For 6 and 20 MeV electrons, after scaling for density $D_{80\%}$ was within 1.3 and 1.5 mm of water, respectively. The high transparency and mechanical flexibility of polymeric gels indicate potential for use as a radiotherapy bolus; differences in density from water may be managed via either using "water equivalent thickness" or by incorporating fused silica into the material.

Keywords

3-D conformal radiotherapy, external beam radiation therapy, bolus, dosimetry, surface dose

Abbreviations

3-D, three-dimensional; CT, computed tomography; HU, Hounsfield unit; PDD, percent depth dose.

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Introduction

Medical errors continue to be problematic within radiation oncology, with high profile errors and misadministrations in highlighting the need for improved and more effective safety measures within this specialty.[1,2] Meeting these demands has been the focus of recent safety initiatives by professional, national, and international organizations.[2-4] One key factor to safe and effective radiation therapy is accurate and reproducible alignment of the patient anatomy prior to delivering the radiation.

Patient alignment for radiation therapy is often complex, including specific patient geometry, immobilization devices, and often a bolus, or mass of scattering material placed on the patient's surface which provides additional scattering, build-up, or attenuation of the radiation beam. The bolus material complicates the patient alignment, as it can obscure the localization marks on the patient surface. In addition, accurate dose often requires that no gaps be present between the skin and bolus, which is difficult to verify with an opaque or translucent

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bolus material. Volumetric image guidance can assist in patient localization but is not always available and is often not utilized on a daily basis, especially for electron treatments, which often require a bolus. Indeed, many electron treatments do not involve a planning computed tomography (CT) and instead rely entirely on a visual "clinical setup" of the patient anatomy within the room. Furthermore, though many potential risks have been mitigated using software solutions such as record and verify systems,[5] these systems cannot replace the visual verification of the treatment alignment by the therapists within the room. Thus, transparency is a quality of a bolus material that could help facilitate accurate and reproducible alignment of patient anatomy.

Numerous improvised and commercial solutions are used or have been proposed as radiotherapy bolus, including liquid water held within a container,[6] wet gauze,[7] elastomer materials[8] moldable waxes and opaque dough,[9,10] custom milled or moldable wax,[11] thin solid or mesh thermoplastic material,[12] a commercial water-based gel material typically used for superficial burns,[13] and preformed gel sheets.[12] Although many of the currently used bolus materials are translucent, there are limited options that are transparent, especially for cases requiring a relatively thick bolus. In addition to transparency, the ideal characteristics of a bolus material include tissue equivalence, sufficient flexibility to conform to surface contours, durability with use and radiation dose, as well as being easy to clean, and nontoxic.[8]

Polymeric gels consist of a high-molecular-weight elastomeric network with plasticizer filling the interstitial spaces of the network. The network of long polymer molecules imparts elasticity to the material and constrains the plasticizer, which in turn imparts flexibility.[14-16] Polymeric gels vary widely in scope, as their composition can be varied to achieve widely varying hardness, elasticity, transparency, and density.[16,17] These materials have been applied to a multitude of purposes including pharmaceuticals,[18,19] art conservation,[20] and food applications[21] but have not yet been applied as a radiation therapy bolus. Some advantages that polymer gels would provide as a radiotherapy bolus include being flexible, odorless, biologically nontoxic (avoiding potentially questionable plasticizers such as phthalates[22]), and its potential to be highly transparent. Here, we investigate the use of oil gel-based materials as a radiotherapy bolus, including a survey of relevant physical and dosimetric properties.

Materials and Methods

Polymeric gels of varying combinations of a plasticizing food-grade mineral oil and 3 different hydrogenated styrenic block copolymers were manufactured with physical properties that are desirable as a radiotherapy bolus. To manufacture the polymeric gels, the mineral oil was heated to 140°C and one of the 3 block copolymers (labeled polymers A, B, and C) was added to the mineral oil during constant stirring. The polymers were obtained from KRATON (Houston, Texas), the details of which are provided in Table 1. The polymer consisted of 8% to 20% of the mixture by weight. Most polymer gel combinations consisted only of mineral oil and block copolymer; however, we also experimented with adding particulate-fused silica to increase the density and make the radiological properties of the bolus material more similar to water. Mechanical properties of these polymeric gel materials were quantified including maximum tensile stress, maximum tensile strain, Young modulus, and contact angle. For the polymeric gel deemed to have the best fit as a radiotherapy bolus, additional physical properties were quantified including mass and electron density, optical transparency, and clinical aspects such as durability after repeated irradiation and cleaning. Dosimetric quantities were also quantified including relative dose curves as a function of depth in the medium for clinical electron and photon beams. Mechanical properties of a commercial radiotherapy bolus material (Superflab, Mick Radio-Nuclear Instruments, Mount Vernon, New York) were also measured for comparison.

Table 1. Vendor-Specified Details for Polymers Used in the Study.[a]

Property	Polymer A	Polymer B	Polymer C
Copolymer type	Linear	Linear triblock	Linear triblock
Content	Styrene and ethylene/ butylene	Styrene and ethylene/ butylene	Styrene and ethylene/ butylene
Specific gravity, g/cc	0.91	0.91	0.91
Styrene/rubber ratio	33/67	30/70	30/70
Viscosity (cP), 5% wt	42700	12	18
Tear strength, J/m	475	21	75
Shore A hardness	60	69	72

[a]Polymers were obtained from KRATON.

Mechanical Characterization

Tensile properties of the polymeric gel materials (maximum stress, strain, and Young's modulus) were evaluated because they help define its suitability for routine clinical use; an optimal bolus material will be flexible enough to conform to the anatomical contour but also strong enough so as to not break during routine clinical use. Some measures that help to define these ideal characteristics include maximum strain, tensile stress, and Young modulus. To measure these characteristics, 1-inch wide sections of each material were cut and attached to a rheometer (Tinius-Olson, Horsham Pennsylvania). The initial length and cross-sectional area were measured. The tensile force, σ, was measured until fracture, and the force and extension data were converted to stress in gigapascals (GPas) by the equation:

$$[\sigma = F/A] \tag{1}$$

where F is the force applied on area A. Strain, ϵ, is unitless and was obtained as the change in length L by the equation:

$$[\varepsilon = \Delta L/L_0] \tag{2}$$

where ΔL is the change in length and L_0 is the original length of the material. Plots of stress versus strain were generated. The maximum strain was calculated by taking the largest extension of the material before fracture and dividing this value by the original length. The tensile stress was calculated by taking the stress of the material at fracture. To calculate Young modulus, E, the slope of the stress–strain curve was measured at "small" strain values, according to the equation

$$[E = \sigma/\varepsilon] \tag{3}$$

which provides a result in pascals [16]. "Small" strain refers to lengths in which there is linear response in the material and is in most case before the strain equals 0.1. There is a range of strain values for the Superflab due to the method of calculating Young modulus. The low value is calculated at small strains (below 0.5 strain), and the high value was calculated when the stress–strain curve was linear (at strains ranging from 1 to 1.5).

Mass and Electron Density

Mass density was calculated via gross measurement of mass and volume as well as using a conversion from CT Hounsfield units (HUs). A CT scan was acquired using 120 kVp, using a standard clinical CT scanner (Biograph mCT, Siemens Healthcare, Erlangen, Germany). A calibration curve to convert between HUs and mass and electron density was obtained using a CT of a calibration phantom (Electron Density Phantom Model 062, CIRS, Norfolk Virginia) with the same acquisition settings. Hounsfield units of the polymeric gel materials and phantom inserts were measured in the central axial CT slice using a 1 × 1 cm^2 region of interest.

Effective Atomic Number (Z_{eff}) and Dosimetric Characteristics

The effective atomic number (Z_{eff}) of the polymeric gel material was also calculated as a function of photon energy from the raw materials and their proportions using the auto Z_{eff} software.[23] We measured the percent depth dose (PDD) of 1 sample polymeric gel material and compared it to the same curve achieved within a water equivalent plastic phantom. This measurement was carried out with a plane parallel chamber (PPC05, IBA dosimetry) at a source-to-surface distance of 100 cm, with a 7 × 7 cm^2 field size, as seen in Figure 1A. We utilized a 7 × 7 cm^2 field rather than a standard 10 × 10 cm^2 field due to the size of the bolus created, as the 10 × 10 cm^2 field would extend near the edge of the bolus especially at larger depths. The radiation source was a 6 MV photon beam from a Varian 600C linear accelerator (Varian Medical Systems, Palo Alto, California). We also measured the PDD using 2 clinical electron beams, 6 and 20 MeV, which represent 2 extremes of the electron energies used clinically in our radiation therapy department. For measuring the electron PDD curves, the measurements were made at 100 cm source to surface distance with a 10 × 10 cm^2 applicator and the nominal open, square (10 × 10 cm^2) cutout, with the plane parallel chamber (illustrated in Figure 1B).

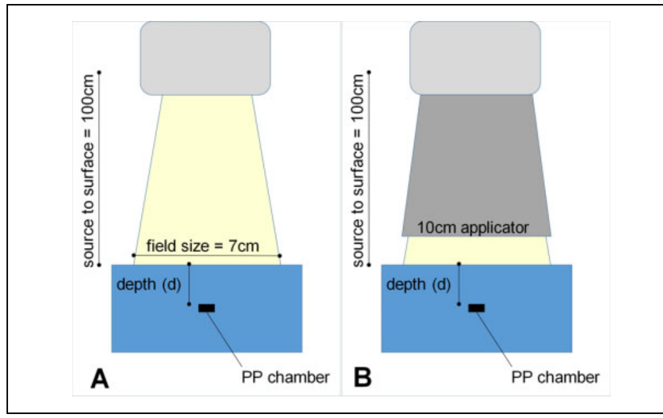

Figure 1. Experimental setup for measurement of percent depth dose in solid water and polymeric gel material for photons (A) and electrons (B).

Results

Mechanical/Tensile Properties

Figure 2 shows the stress–strain relationship for 7 polymer gel combinations, as well as the commercial bolus material. This highlights the wide variety of tensile properties that can be achieved using polymeric gels; the clinical bolus material fits within this range. Table 2 shows the maximal strain, tensile stress, and Young modulus of these same materials. Since an ideal characteristic of bolus is the ability to conform it to varying patient contours, a soft, less stiff material is desired, likely with a modulus similar to the commercial bolus product in the range of 0.035 to 0.050 GPa. Plotted in Figure 3 is the effect of polymer concentration (polymer B from Figure 2 and Table 2) on Young modulus and tensile strength for an example oil gel formulation; the mechanical properties can be adjusted to achieve a desired material hardness using the polymer concentration.

The polymeric gel consisting of 15% polymer B by weight from Table 2/Figure 2 was selected for further (radiological) characterization, as it was one of the materials with mechanical properties that were most amenable for use as a radiotherapy bolus.

Mass and Electron Density

Mass and electron density were measured for the selected polymer gel material (polymeric gel with 15% polymer B); mass density as measured by weight/volume measurement was 872

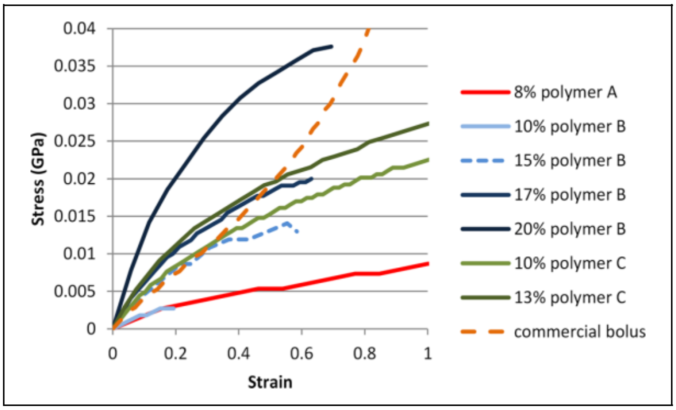

Figure 2. Stress–strain curve measured for polymeric gel combinations, compared to a commercial bolus material. The polymer gel with 15% of polymer B was selected for further characterization.

Table 2. Tensile Properties for Polymeric Gel Materials.

Polymeric Gel Combination	Max Strain, Unitless	Tensile Stress, GPa	Young Modulus, GPa
8% Polymer A	3.5 (no fracture)	0.022 @ 3.5 strain (no fracture)	0.015
10% Polymer B	0.21	0.0027	0.021
15% Polymer B	0.62	0.013	0.039
17% Polymer B	0.65	0.020	0.066
20% Polymer B	0.75	0.038	0.12
10% Polymer C	1.2	0.026	0.052
13% Polymer C	1.2	0.029	0.071
Commercial material	1.6	0.12	0.037-0.097

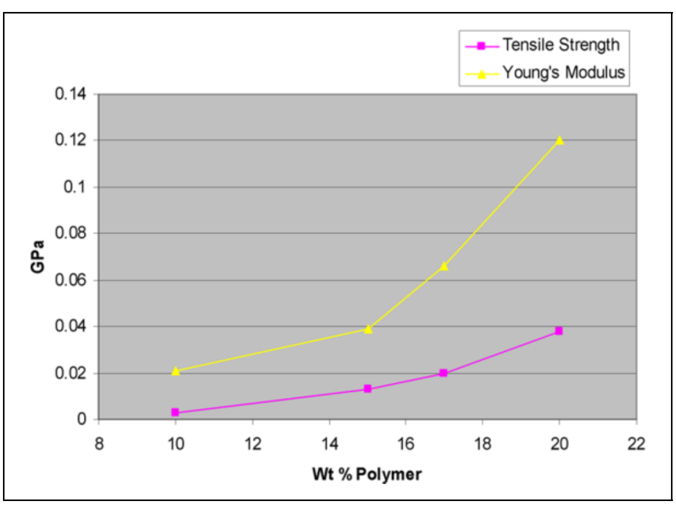

Figure 3. Tensile strength and Young modulus as a function of percent weight of polymer B. Mechanical properties of the bolus material can be optimized by adjusting the polymer content.

$\pm$ 12 g/cm^3, using CT HU, the mass density was 896 $\pm$ 13 g/cm^3, with the corresponding electron density being 2.95 $\pm$ 0.04 $\times$ 10^{23} electrons/cm^3. As this density is slightly lower than water, we also investigated introducing silica into the oil/polymer combination to increase the density. As silica has a density of 220 g/cm^3, including 9% by weight gives the bolus a mass density equal to water. The atomic composition of these bolus materials is given in Table 3.

Table 3. Atomic Composition of Finalist Bolus Material (15% Polymer B Prior to Silica).

	Mass Fraction	
Element	Bolus (no silica)	Bolus (9% silica)
H	0.151	0.137
C	0.849	0.773
O	0.000	0.048
Si	0.000	0.042

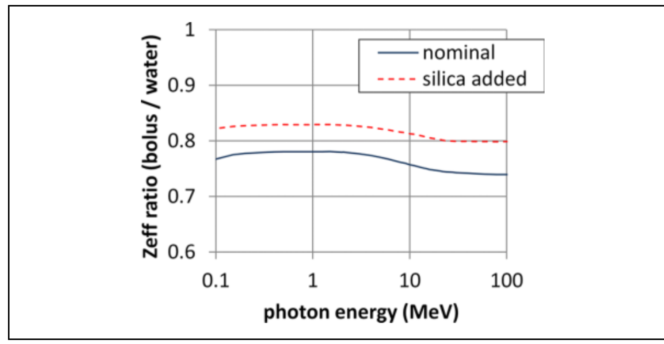

Figure 4. Effective atomic number as a function of photon energy for the finalist bolus material (15% polymer B prior to silica) with and without added silica (9% by weight) to achieve equal density to water.

Effective Atomic Number (Z_{eff}) and Dosimetric Characteristics

Figure 4 shows the effective atomic number (Z_{eff}) as a function of photon energy for the selected polymer gel material (polymeric gel with 15% polymer B) with no silica added and with sufficient silica added to increase the mass density to that of water (9% by weight). The polymeric gel bolus has a slightly lower effective atomic number than water, with a nearly constant ratio in the therapeutic energy range. The relative depth dose curve is shown in Figure 5 for the sample bolus (no silica) and for solid water in a clinical 6MV photon beam (A), and clinical 6 MeV (B) and 20 MeV electron beams. Also shown is the PDD curve for the bolus when the depth is scaled by the ratio of its density (from CT) and that of water (0.896). The curves for bolus and water coincide after scaling by the bolus density. The PDD for 6MV photons was within 2% of water within the first 2.5 cm and after scaling by the density coincided within 1% out to greater than 7 cm. For the clinical electron beams, a similar scaling could be applied based on density so as to align the PDDs; for 6 MeV, the depth at which the dose curve fell to 80% of its maximum ($D_{80\%}$) was 2.13 cm

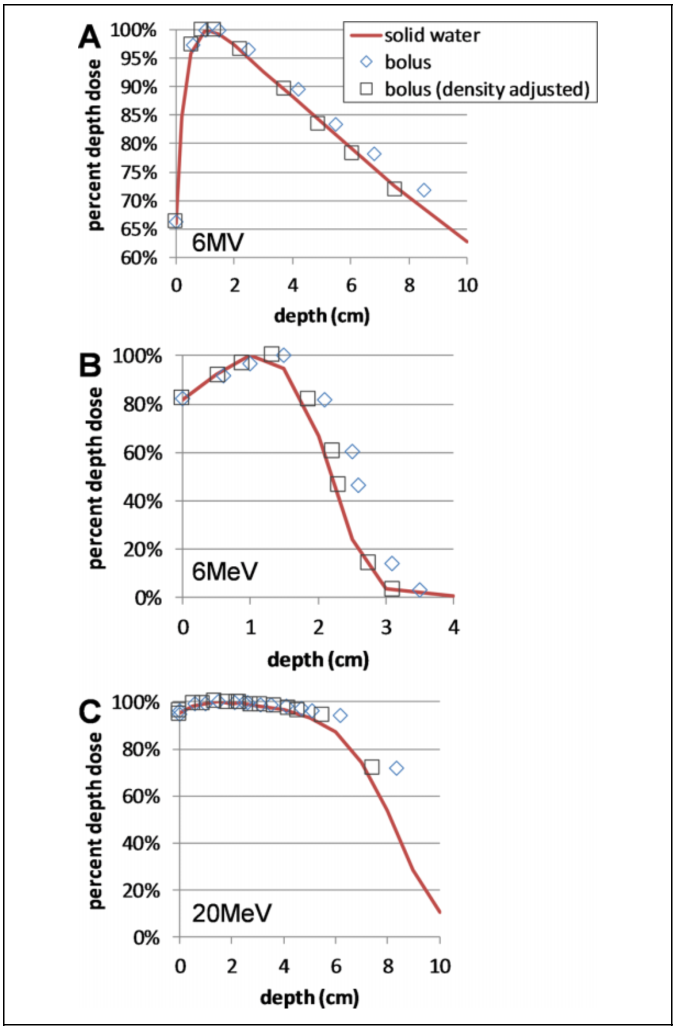

Figure 5. Percent depth dose measured through selected polymeric gel material (15% polymer B, no silica) compared to water equivalent plastic phantom. The dose curves coincide after scaling *x*-axis for density for 6MV photons (A), 6MeV electrons (B), and 20MeV electrons (C).

for the sample bolus compared to 1.76 cm for the solid water; once scaled by density, the difference between these values decreased to 1.3 mm. Similarly for 20 MeV, $D_{80\%}$ was 7.60 cm for the bolus and 6.55 cm for the solid water, but the values were within 1.5 mm after scaling for density.

Optical Transparency

Figure 6 illustrates the optical transparency for the finalist polymeric gel with no added silica compared to a commercial bolus product (A); (B) shows the visibility of fine detail including the text through the material, also shown is the visibility through a 1 cm thick oil/polymer bolus of setup marks under normal room lighting conditions (C) and of the lasers, crosshairs, and light field (D). The visibility was quantified in that 12-point newsprint could be read through 2 cm of material by a person with 20-20 vision.

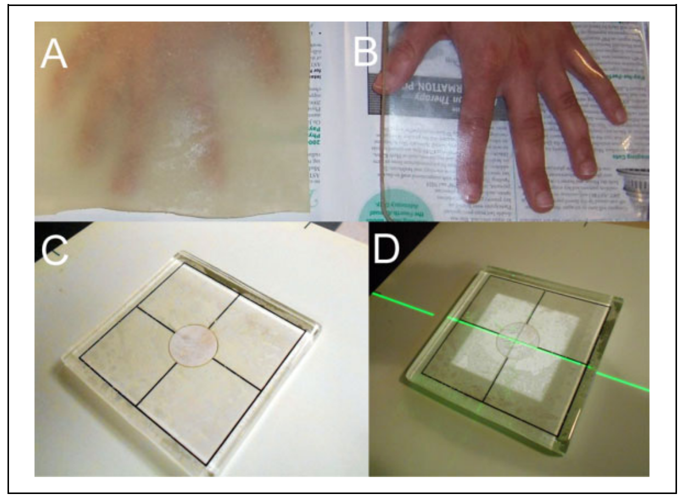

Figure 6. Visibility through commercial bolus (A) and polymeric gel (B-D) materials. Localization marks and the light field could be easily visualized on the surface through the polymeric gel material.

Long-Term Durability

No sign of changes in physical properties was observed after repeated use and cleaning with soap and water or after long-term observation (4 years). Extensive irradiation (1 kGy) also had no observable effect.

Discussion

In this study, we have investigated the feasibility of using polymeric gels for use as a radiotherapy bolus. The main advantage of polymeric gels is that it can potentially be highly transparent. The benefit for day-to-day radiation therapy would be subtle, but self-evident. Many treatments that utilize a bolus, such as many electron radiotherapy cases, do not utilize image guidance technologies and hence depend on the visual setup of the treatment field within the room. Thus, any improvement in the ability to visualize the treatment field through the bolus will be beneficial.

One of the main challenges of polymeric gels as a bolus is the slightly lower density when compared to water. However, we demonstrate 2 potential methods for counteracting this: (1) scaling the thickness of the bolus according to the ratio of its density and water, so as to use a "water equivalent" thickness or (2) including fused silica to increase the density. The most common purpose of a radiotherapy bolus is to increase surface dose, which does not necessarily require a water equivalent material. For electron radiotherapy, water equivalence is useful because the range of the electrons in tissue is highly dependent on the thickness of the bolus, and having a water equivalent material allows the clinicians to easily account for what effect each thickness of bolus will have on the clinical situation without performing a dose calculation. In cases where water equivalence is important, the required thickness is often on the order of 0.5 to 2 cm thick, and the near water equivalence we show here demonstrates that the clinicians can assume a water equivalence for clinically relevant depths of bolus (0.5-2.0 cm) with minimal dosimetric consequence (<2%) for photons or an effective thickness can be used that is equivalent to water with essentially no dosimetric difference (aka, utilizing 1.1 cm thick instead of 1.0).

In this study, we utilized a plane parallel chamber within solid water. This choice was made because we considered it to be more accurate at shallow depths than other potential measures. Also, depth within solid water is easier to define than in water, especially for very small depths. By using the same ion chamber for both the test (bolus) and reference (solid water) conditions, we can negate any potential differences due to P_{fluence}, P_{gradient}, and so on. It should be noted that these measurements may likely systematically overestimate the actual surface dose.[24] Hence, the reported values here should be considered jointly with this potential systematic uncertainty in the measurements.

There have been many other proposed bolus materials.[6-13] The material proposed here would most resemble standard preformed gel sheets,[12] which typically have the advantages of coming in standard thicknesses, deforming to patient anatomy, being radiologically equivalent to water, and being translucent. In contrast, some bolus materials are initially molded to the patient anatomy and then remain rigid,[9-12] which may better conform to the anatomy and avoid potential air gaps, despite their lack of transparency.

Conclusion

In this study, we have investigated the feasibility of polymeric gels for a radiotherapy bolus. Potential advantages of these gels as a radiotherapy bolus include potentially being highly transparent, and the wide range in tensile properties that can be achieved (including properties amenable to a radiotherapy bolus such as being flexible and durable). One challenge identified is the slightly lower density relative to water; however, differences in attenuation of megavoltage photon beams can be assumed to be water equivalent with minimal error after scaling for differences in density, while clinical electron beams could potentially scale the bolus thickness so as to achieve the intended "water equivalent" thickness. In addition, fused silica may be added to the polymeric gel to further diminish the attenuation differences from water. These characteristics indicate the potential that clear polymeric gels have for use as a radiotherapy bolus.

Declaration of Conflicting Interests

The author(s) declared the following potential conflicts of interest with respect to the research, authorship, and/or publication of this article: Cooney, Demehri, Stalnecker, and Kirkpatrick originally filed a patent application relating to use of polymeric gel as a radiotherapy bolus material.

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