Heated Diagnostic Radiology Exam Table

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Abstract

Clinical X-ray examinations sometimes require patients to remain still for long periods of time. A common patient complaint is that X-ray examination tables are uncomfortable; specifically they are too hard and too cold. Patient discomfort is undesirable as an uncomfortable patient is more prone to movement during a procedure, causing poor quality images and complicating diagnosis. The objective of the client and our team is to create a device to ensure patient comfort while preserving radiolucency and patient safety. The focus of our work is to create a device to modify the current surface of the X-ray table through the addition of padding and heat. The designed device consists of polyimide Kapton® film placed between two additional Kapton® sheets and polyethylene foam. To produce heat Dupont™ 5025 MCM Silver Paste® bus bars are used to uniformly run current across the surface of the Kapton®. A microcontroller based control unit powers the device on and off to maintain a chosen target temperature. Finally, the entire device is enclosed in a sterilizable Naugahyde® cover. The materials used, in addition to the original X-ray exam table, do not attenuate more than allowed by CFR-Federal Code of Regulations Title 21.

Introduction/Background

Problem Statement and Motivation

Every year approximately 465,000 X-ray procedures are repeated because of poor image quality caused by patient movement (Weatherburn, 1999). These unnecessary procedures decrease patient satisfaction, take up staff time, and tie up equipment. Moreover, repeat X-rays due to patient movement collectively cost approximately $12.4 million annually (Alanen, 1999).

A common patient complaint is that X-ray examination tables are uncomfortable, i.e. they are cold and hard. Providing a comfortable exam experience for a patient is not only important from a patient satisfaction standpoint, but also from a patient compliance standpoint. During procedures a patient may be more prone to move if uncomfortable. Patient movement during imaging does not allow proper image acquisition.

Market

Currently, there are no products on the U.S. market capable of heating a patient during an X-ray examination. In the U.S. alone, there are over 5,815 registered hospitals (not including health clinics) that have at least one X ray examination table (American Hospital Association, 2009). Annually, in the United States there are over 90.6 million X-Ray procedures performed which our device could be used in (Bhargavan and Sunshine, 2005; and U.S. Census Bureau, 2001). Combined with dentistry, sports medicine, and veterinary fields, this untapped market is massive.

Current Examination Table

The examination table we are working with is shown below in Figure 1.

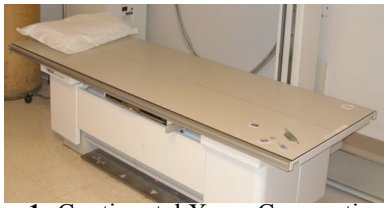

Figure 1: Continental X-ray Corporation "Classic Elevating 4-Way Float Top Table". (www.advanceimaging.net)

Other manufacturers of tables use a design similar to that shown above. The table has a hard laminate surface where the patient lies during a procedure.

The top of the table is 2.2 m long by 0.8 m wide. The tabletop is capable of moving 1.14 m vertically and 0.25 m transversely. The height of the table top is adjustable from 0.55 m at its lowest to 0.84 m at its highest.

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The maximum patient weight as specified by the manufacturer is 158 kg. The standby heat load of the table is 800 BTU per hour.

Design Considerations

The application of our device for use in X-ray or radiology procedures imposes several limitations on our design. The most important constraint on our design is radiolucency. Our device must not attenuate the X-ray beam more than 4.49 percent – the equivalent of 1 mm of Aluminum (Hubbell and Seltzer, 2009). It also must not introduce artifacts into the X-ray image.

Patient safety is also of high concern; the possibility of burns or electrocution is not acceptable. To prevent patient harm during direct skin contact, the device cover must be non-toxic, anti-microbial, and easily sterilizable in the event of bodily fluid or pathogen exposure.

The device must respond rapidly to a user selected temperature setting. The amount of heat transferred to the patient should be adjustable within 1 degree Celsius and must never exceed 44 degrees °C, the temperature at which skin burns begin to occur (Exponent, 2008).

Lastly, the device needs to afford some level of increased comfort through cushioning without causing anatomical distortion of the patient.

Materials

Kapton RS® Polyimide Film

Kapton RS® films, manufactured and distributed by DuPont™, can withstand high temperatures (up to 350 degrees Celsius) making them exceptionally useful as flat heaters. Since Kapton® is a polyimide film completely comprised of low molecular weight organic material. It is radiolucent and ideal for X-ray imaging.

Kapton® is a 50.8 μm (2 mil) thick, bilayer polyimide film. One layer of the film is 25.4 μm (~ 1 mil) thick and uniformly loaded with conducting polymers, producing a tightly controllable bulk resistance. The standard resistance of the conductive side of Kapton® is 100 ohms/square. The other layer of the Kapton® film is also .0254 mm (~1 mil) thick and has a dielectric strength of 2,500 volts. This layer acts as a strong electrical insulator. As the conducting and insulating characteristics are distributed throughout the bulk of the Kapton® material – rather than coated on the surface – they cannot be cracked, rubbed off or damaged during normal use.

Polyethylene Fine Cell Padding

The fine-cell polyethylene pad is 3.2 mm thick – enough to make the X-ray exam surface soft and comfortable, but not so much as to distort the anatomy of the patient.

Naugahyde Cover

A medical-grade Naugahyde cover (Mosehart-Schleeter, Texas) encases all layers of the heating pad. The Naugahyde is water and oil resistant, antifungal, and anitmicrobial. Also, it can withstand common sterilization solutions (e.g. bleach or ethanol) for up to 500,000 wipe down cycles.

Final Prototype

Heating Pad

The prototype consists of four distinct layers, each measuring 60 inches long by 24 inches wide (Figure 2). These layers, from the table surface to the patient, are: fine cell polyethylene pad, followed by three layers of Kapton®.

The bottom layer, or the layer closest to the table, is fine cell polyethylene padding. The purpose of this padding is for patient comfort. The compliant nature of the soft polyethylene foam provides a cushioned surface for the patient to lie on. The padding also provides thermal insulation.

The second and fourth layers, each consisting of a 50.8 μm thick Kapton® sheet, serve as an electrical insulator and safety mechanism. The Kapton® is oriented such that the insulating, dielectric side of the film is touching the heating layer of Kapton® film. The conductive side is electrically connected to earth ground, ensuring that if any charge or current were to make it past the insulating layer, it would be directly

dissipated. The orientation and placement of this layer is a safety mechanism to protect against patient electrocution.

The third layer (also Kapton®) generates the heat. Bus bars, made from DuPont™ MCM 5025 Silver Paste® run the entire length of the film. By connecting the bus bars to an AC voltage source, electrical current is passed across the width and through the bulk of the sheet. As current flows through the bulk of the sheet, the sheet behaves as a uniform ohmic heater. This heat then travels to the patient primarily through convective flow.

Finally, all four layers are encased in the Naugahyde sheath. In addition to providing a small amount of electrical insulation, the Naugahyde surface can be easily sterilized between patient procedures. It also has the added benefit of being comfortable and aesthetically pleasing. This final product, the sheath and the four layers it contains, is collectively referred to as "the pad".

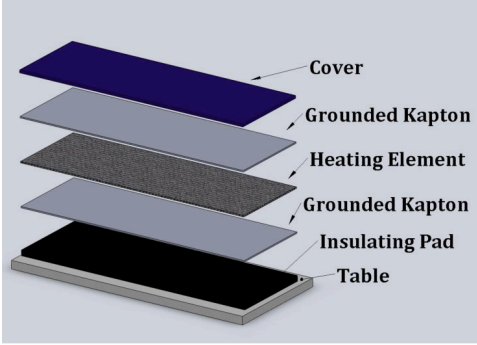

Figure 2: Schematic of the pad.

Power is supplied to the 2pad from a 120 volt AC wall outlet. The power line includes a ground fault interrupter (GFI) plug. If the current leaks into one of the safety grounded, insulating sheets of Kapton®, the GFI will trip and power will be cut to the pad. The 120 volts applied to the 35 ohm resistance heating element results in a current of roughly 3.5 amps flowing through the heating Kapton® layer.

Temperature Controlling Circuit

In order to incorporate a degree of patient and technician control, a circuit was included in the final design. The entire circuit is controlled by two ATMega8 microcontrollers, which receive several inputs in order to control heat output by the pad.

Each microcontroller controls an alphanumeric LED display; one displaying the temperature set by the technician, the other displaying the actual temperature at the surface of the pad. The technician changes the set temperature by pushing one of two buttons, thereby increasing or decreasing the set point, which feed directly into both microcontrollers. The set temperature is displayed by an LED display. The pad temperature is monitored by three temperature sensors attached to the second microcontroller. These sensors send information to the microcontrollers, allowing for control of heat, as well as display of the actual pad temperature (in relation to the set point).

Current flow, and therefore heat generation, is controlled by a relay, which is attached to the same microcontroller that is monitoring the pad temperature. If the pad temperature is below the set temperature, the relay is turned on, allowing current to flow to the pad. Current flows to the pad until the actual temperature exceeds the set temperature by two degrees. At this point, the relay is shut off, effectively stopping current flow. Once the temperature is two degrees below the set temperature, the relay is turned back on, resuming current flow to the pad. It is this on-off action that keeps the temperature at a relatively steady state.

If, at any point, the temperature exceeds 44 °C, the relay immediately shuts off and current flow is stopped. This safety mechanism prevents the technician from setting the temperature too high, and ensures that the pad never becomes hot enough to burn the patient.

Results

To validate the device for clinical use, several steps were taken to ensure that it behaved as intended and did not substantially affect X-ray image quality.

Radiolucency

If the device attenuated the X-ray signal too much, or introduced artifacts into the image, the design would not be practical for clinical application. Therefore, it was important to prove that the radiolucency of the device was sufficient.

First, the device was qualitatively investigated. A multipurpose chest phantom (N1, Kyoto Kagaku Co., Tokoyo, Japan) was used to simulate X-ray examination of a human thorax. The pad was positioned on an X-ray table and the phantom placed on top of the pad. Following standard operating procedure, X-ray images were taken at 70 kV, 40 mAs, and 66.8 ms. Typical results can be seen in Figure 3.

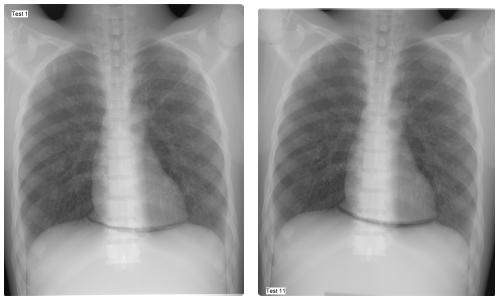

Figure 3: Comparison of radiographs of a human chest phantom. At left is an image acquired while using the device, at right is an image without the pad. Qualitatively, the two images differ negligibly.

The radiographs were examined by a trained expert and deemed suitable for diagnosis. The pad did not introduce artifacts or noticeably attenuate the X-ray beam. Importantly, the fine features of the phantom (such as the aveoli of the lungs) were not distorted by the device, demonstrating its utility in procedures where a high level of precision is required.

Next, to assess the performance of the device quantitatively, we looked at the attenuation of the pad compared to air alone. To test this, the pad was placed directly over half the X-ray film cartridge. The other half was left uncovered and the whole cartridge exposed at 70 kV, 1 mAs and 2.0 ms according to standard operating procedure. A radiograph is displayed below in Figure 4.

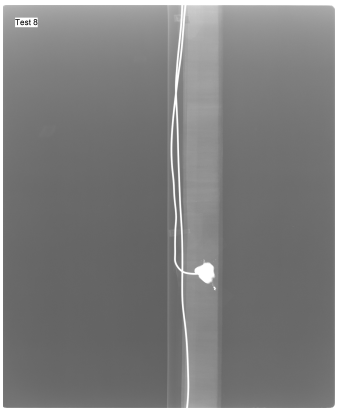

Figure 4: Radiograph comparison of device (at right) and air (left). The light band separating the two fields is the bus bar/epoxy/wire combination and was not included in the attenuation calculations.

After the radiograph was obtained, it was processed using ImageJ (NIH, Bethesda, Maryland) and a custom plug-in (Fuji_Transform, Prof. Wally Peppler). With data obtained during imaging, the software was used to obtain exposure values at each pixel. This information was then averaged over the area of the pad-covered and the uncovered halves of the image.

It was found that the pad attenuated, on average, 1.11 percent more than air. This is well below the 4.49 percent allowed by federal regulation. Moreover, the standard deviations of the attenuation values for the device and air halves closely matched each other. This means that the X-ray permeability of the pad is similar to that of air in terms of uniformity.

Heating

It is necessary to ensure that the current supplied to the pad actually produces heat that reaches the patient. Moreover, it is important that the heat is evenly distributed across the pad. Specifically, differences in resistance at the bus bars, especially due to the deposition method, could cause localized hot spots. These uneven heating patterns would not only jeopardize patient comfort, but could also thermally stress the bus bar connection and cause further fluctuations in resistance and eventually failure.

To test the uniformity of the heat field, a Ti25 Thermal Imager (Fluke, Everett, WA) was used.

First, the heat gradients produced during the heating of the pad were examined. With target temperature set to 46 °C, the pad was allowed to heat up. Measurements were taken every ten seconds over the ninety seconds it took for the pad to reach the target temperature and every thirty seconds as the pad cooled to room temperature. A representative image is shown below in Figure 5.

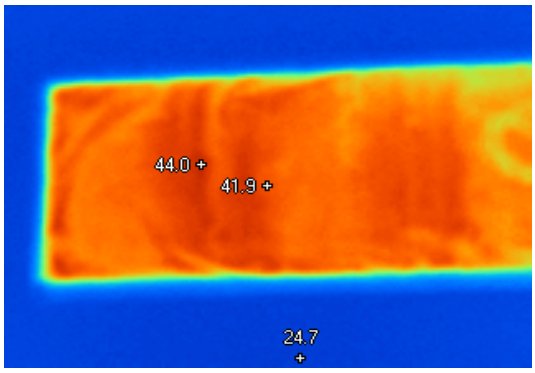

Figure 5: Image of device at t = 180 seconds. The variability across the pad is due to creases in the cover and air pockets. The applied weight of a patient is expected to eliminate these subtle nonuniformities.

These data qualitatively demonstrated the uniform heating of the device. To quantitatively prove that the temperature gradients are sufficiently small, values of each individual pixel were interrogated. After discarding ambient data, a histogram (Figure 6) was constructed and statistically analyzed.

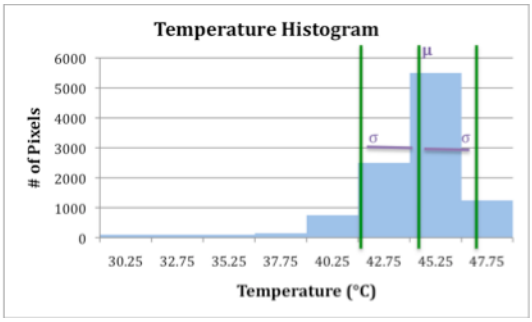

Figure 6: A temperature histogram of the device at t=90 seconds. Approximately 80 percent of the data points are within one standard deviation of the mean, indicating a consistent temperature across the pad.

While the standard deviation was larger than expected, it was nevertheless within acceptable limits. The histogram demonstrates that eighty percent of the pixel values are within one standard deviation of the mean. Moreover, Figures 5 and 6 indicate that no hot spots form along the bus bars during heating.

Finally, the response time was tested. The target temperature was set at 46 °C and the temperature of the pad was taken every ten seconds as it rose from room temperature. The results are shown below in Figure 7.

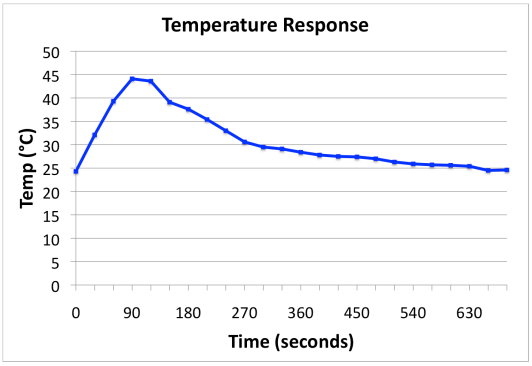

Figure 7: Temperature vs. time as the pad heats up from room temperature. At t=95, the target temperature was reached and power was turned off. The pad was then allowed to cool back to ambient temperature.

Our device reached the target temperature in just over 90 seconds. For clinical applications, this means it should reach temperatures of ~37 °C in under a minute.

Conclusion

By applying new materials to the problem of radiolucent heating, we were able to create a clinically useful device for ensuring patient comfort without sacrificing X-ray image quality. The device, which relies on a Kapton heating element, creates an even heating field and does not introduce artifacts. The entire device only attenuates the beam 1.1% more than air, well within federal guidelines. Altogether, the device presents a simple, attractive opportunity to increase patient compliance and comfort in clinical settings without sacrificing functionality.

Works Cited

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